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Biomechanical performance and relevant mechanism of physical medicine and rehabilitation for neuromusculoskeletal disorders

Edited by
Qipeng Song, Li Li, Pui Wah Kong, Lin Wang and Feng Wei

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Biomechanical performance and relevant mechanism of physical medicine and rehabilitation for neuromusculoskeletal disorders

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Editorial: Biomechanical performance and relevant mechanism of physical medicine and rehabilitation for neuromusculoskeletal disorders

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Editorial on the Research Topic
Biomechanical performance and relevant mechanism of physical medicine and rehabilitation for neuromusculoskeletal disorders

Introduction

Biomechanics plays a crucial role in evaluating the effectiveness of physical medicine and rehabilitation for neuromusculoskeletal disorders. Assessments can be utilized for conditions such as degenerative dysfunction (e.g., falls or knee osteoarthritis in older adults) and sports-related injuries (e.g., ankle sprains or anterior cruciate ligament (ACL) injuries). Movements and daily functional activities of those who suffer injuries can be contrasted with their pre-injury condition or uninjured individuals. Pioneering investigations have taken a step forward by leveraging biomechanical metrics to formulate strategies within physical medicine and rehabilitation and delve into the underlying efficacy mechanisms (e.g., Li et al., 2019). Nonetheless, such inquiries remain somewhat scarce—this Research Topic aimed to stimulate the dissemination of more pertinent endeavors in this area.

The editorial team orchestrated this Research Topic to compile the advancements that elucidate the progress in physical medicine and rehabilitation for neuromusculoskeletal disorders. This Research Topic, "*Biomechanical Performance and Relevant Mechanism of Physical Medicine and Rehabilitation for Neuromusculoskeletal Disorders*," encompassed 31 studies, comprising 24 original and seven review articles.

Mechanisms of sports-related injuries

In sports-related injuries, the primary objective of injury prevention lies in minimizing injury risks and understanding the underlying mechanisms, particularly at knee and ankle joints. Among individuals who underwent anterior cruciate ligament reconstruction (ACLR), diminished postural control capability is a significant obstacle in their return to competitive activities (Wang et al.). A noteworthy observation was the reduced dynamic postural control exhibited by these participants, coupled with noticeable deficiencies in proprioception. Cui and group compared the surface electromyographic features of lower limb muscles in people with ACL injuries at different stages: 6 months before the injury, 6 months and 1 year after ACLR (Cui et al.). Their study could help determine which muscles require more training and which exercises are best suited for interventions. Based on the reports of retrospective studies, Grodman et al. observed that levels of activities or maneuvers during non-contact conditions among ACL-injured patients significantly increased 6 months after their injuries. Their findings provided evidence that changing levels of certain activities or maneuvers may play a role in ACL injury risk. Meanwhile, Long et al. utilized an electromyography (EMG) method to evaluate the risk of knee osteoarthritis following unilateral ACLR (Long et al.). The results showed that the peak tibial compression force, knee flexion and ankle dorsiflexion angles, and the muscle force of the rectus femoris and vastus medialis on the healthy side were greater than that on the surgical side during jogging. This indicated that the body automatically exhibited compensatory mechanisms on the healthy side to reduce the risk of further injury on the surgical side.

The lower extremity joints collectively form a tightly linked kinetic chain; therefore, an injury to one joint might cause compensation in another joints. Xu et al. reported that participants with chronic ankle instability (CAI) altered proximal lower extremity joint motion strategies during lateral cutting, jump landing, and abrupt stopping, potentially elevating the risk of ACL injury (Xu et al.). Kong's group explored the effect of mental fatigue on biomechanical characteristics of lower extremities in people with functional ankle instability (FAI) (Kong et al.). They observed that the ankle stiffness of people with FAI had no significant change during anticipated side-step cutting but presented less ankle stiffness on the injured side during unanticipated side-step cutting to increase ankle instability and the risk of re-injury. Under conditions of visual deprivation, Meng et al. observed that individuals with FAI presented a greater risk of instability in the affected limb compared to the other limb (Meng et al.). The authors recommended that further research to investigate the influence of varying visual conditions on stability and gait performance.

Rehabilitation strategies for sports injuries

This Research Topic included innovative methods and interventions, e.g., kinesio taping, electronic interventions, and machine learning. Li et al. utilized Kinesio Taping (KT) intervention on muscle strength and postural control in collegiate basketball athletes with FAI (Li et al.). They observed that the plantar flexor and dorsiflexor moment increased by 20% and 34%, respectively. Chen et al. explored the effects of KT therapy on gait and EMG in people suffering from stroke, showing that the root mean square of tibialis anterior EMG signals in the hemiplegia limbs decreased during walking after KT treatment (Chen et al.). Que et al. utilized the KT and vibration treatment (VT) to alleviate delayed-onset muscle soreness among college students (Que). They reported that KT and VT could reduce pain and strength loss to varying degrees. VT was better than KT in improving pain, and the combined intervention worked better than single interventions. The EMG-based robots were superior to conventional therapies in improving upper extremity motor control, spasticity, and activity limitation. Huo and coworkers showed that the efficacy of the treatment was better in people with post-stroke who received EMG-based robotic techniques combined with electrical stimulation (Huo et al.).

Virtual reality training improved lower extremity muscle strength and postural control. Wang et al. reported that an 8-week virtual reality training improves the muscle strength of hip flexors and extensors, knee flexors and extensors, and ankle plantar flexors (Wang et al.). However, no significant improvement was observed in postural control ability in adolescents with intellectual disability. Meanwhile, Gao et al. built a support vector machine to recognize the gait characteristics of both lower extremities before and after fatigue to prevent running injury, monitor movements, and assess gaits (Gao et al.). Based on continuous wavelet transform, Bai et al. reported that the impact force characteristics in the walking support period differed between people with and without flatfoot in the time and frequency domains (Bai et al.). Huang et al. showed that the sample entropy in the anteroposterior and mediolateral directions of the corrected vision state was greater than in myopia and eyes-closed conditions (Huang et al.). The maximum flexion angles of ankle and knee joints were in the following order: corrected vision, myopia, and eye-closed. That is, the stability of static and dynamic posture in corrected vision was worse than other vision conditions, and the activation and work of lower extremity muscles were increased. Meanwhile, Zeng et al. recruited 22 male college students with normal visual acuity to test gait characteristics at normal and with 150 ° and 450 ° concave lenses (Zeng et al.). They observed that hyperopic interventions impacted the kinematic gait characteristics in male college students, mainly in terms of altered postural control, increased instability, and increased difficulty in maintaining trunk stability with the risk of injury.

Mechanisms and rehabilitation approaches of fall prevention

Fall prevention in older adults is a critically important Research Topic in the neuromusculoskeletal area, as falls can cause further health issues and lead to functional impairments (Peng et al., 2019). In this area, Wang and coworkers recruited 166 older adults in three age groups (young: 60–69 years, old: 70–79 years, and older: ≥ 80 years) and compared their differences in tactile sensation, proprioception, muscle strength, and postural control (Wang et al.). It was observed that the young and old adults had better postural control, tactile sensation, proprioception, and muscle strength compared to the older adults. Proprioception correlated

with postural control in young and old adults but not in older adults. The worsened proprioception among the older could be the key to increased fall risks. Aging involves a decline in muscle and bone mass and a deterioration in cognitive function, especially during dual-task conditions (Song et al., 2018). Li et al. recruited 15 older adults to evaluate the test-retest reliability of kinematics and kinetics during single-task and dual-task stair walking (Li et al.). They observed that the intraclass correlation coefficient of kinematics and kinetics ranged from fair to excellent. These results may help researchers to access biomechanics of dual-task stair walking in the elderly and to interpret the effect of interventions in this population.

Physical activity plays a crucial role in the wellbeing of elderly individuals. Exercise intervention is an important approach to preventing falls, improving the quality of life, and alleviating pain symptoms in older adults (Song et al., 2018). Zhao et al. adopted typical Tai Chi to improve lower extremity inter-joint coordination and variability in older female adults (Zhao et al.). They observed that the mean absolute relative phases of hip-knee and knee-ankle and the deviation phase of hip-knee were significantly less in older adults who practiced Tai Chi than those who did not. Wang et al. reported the results of an 8-week Chinese herb hot compress combined with therapeutic exercise regarding people with knee osteoarthritis (Wang et al.). The intervention significantly relieved pain and decreased proprioception thresholds of knee extension and ankle plantarflexion. The results also showed improved functional performance, including the Time Up-and-Go test and 20-m walk test, compared to the therapeutic exercise training. Sun et al. described the role and regulation mechanism of Chinese traditional exercises, including Tai Chi, Baduanjin, Wuqinxi, and Yijin Jin, which had different emphases and could improve bone density in various parts of the body (Sun et al.). These exercises were found effective for older adults in improving the bone density of the exercisers and relieving pain, improving postural control, and regulating their psychological state. Moreover, neurological disorders with dyskinesia seriously affect older adults' daily activities and are associated with the degeneration or injury of the musculoskeletal or nervous system (Song et al., 2021). Wu et al. described that most of the reviewed studies reported poor motor performance and higher cortical activation of Parkinson's disease, stroke, and multiple sclerosis in older adults than healthy individuals (Wu et al.). More than 5 weeks of walking training or physiotherapy could promote motor function and cortical activation in people with Parkinson's disease and stroke. Traditional Chinese exercises have been utilized to enhance adults' physical and mental wellbeing (Song et al., 2018). Zhang et al. explored the effect of an 8-week Bafa Wubu of Tai Chi on college students. Anxiety and depression were significantly decreased in the Tai Chi group (Zhang et al.). The authors suggested these benefits could stem from modulating activity in the left middle frontal gyrus and the right middle frontal gyrus of the orbital part, respectively.

Neuromusculoskeletal applications in sports performance

In other aspects of research on the neuromusculoskeletal area, injury prevention has consistently been a focal point in sports biomechanics. A front-view video analysis approach is a relatively safe, cost-effective tool for practitioners to measure foot inversion angles at the initial foot contact. Iskandar et al. further indicated that front-view foot inversion angles at the initial foot contact could be used to determine rearfoot inversion angles when crossover gait obstructs the back camera view (Iskandar et al.). A narrative review by Mei et al. suggested tracking foot motion across segments and converting 2D motion to 3D shape, using Machine Learning to address foot mechanics' nonlinear links to shape or posture (Mei et al.). They suggested standardizing wearable data to quickly predict instant mechanics, load, injury risks, and foot tissue-bone adaptation. They further suggested correlating with shapes and analyzing real-life dynamic shapes and posture through markerless real-time methods for accurately evaluating clinical foot conditions and footwear development. A systematic review by Liu et al. reported the relationship between running economy and lower extremity stiffness, namely, vertical stiffness, leg stiffness, and joint stiffness (Liu et al.). They showed that endurance runners' vertical, leg, and knee stiffness were negatively correlated with running economy. They suggested that maximum oxygen uptake and speed could be used to determine whether the runner could take full advantage of leg stiffness to minimize energy expenditure. Li et al. studied the effects of blood flow restriction training and electrical muscle stimulation during low-intensity squat training (Li et al.). The combined interventions improved the muscle strength of the lower extremities by promoting muscle hypertrophy and improving muscle activation. The authors suggested the results were likely due to such a combination compensated for the limitations and deficiencies of the two intervention methods when applied alone.

Moreover, Keriven et al. explored the effect of peripheral electromagnetic stimulation after an eccentric exercise-induced delayed-onset muscle soreness protocol in professional soccer players (Keriven et al.). They observed that stimulation did not significantly improve the lower extremities' power and strength but decreased the vastus medialis' peripheral sensitization after the eccentric exercise protocol. Female college students' body postural control ability improved substantially after comprehensive sports activities (Zhang et al.). The participants had a more rational body composition and continued natural bone mineral density increase, unrelated to exercise intervention. Regarding athletic performance, Liang et al. reported that the lower extremity joint range of motion and stiffness significantly impacted the foot-up serve performance in tennis players (Liang et al.). A greater joint mobility and lower-limb stiffness promoted better performance during the foot-up serve preparation stage. For methodological considerations, Montoro-Bombú et al. provided information on criteria for determining the volume and intensity of drop jumps during plyometric training programs to help researchers conduct new plyometric training programs (Montoro-Bombú et al.).

This Research Topic aimed to encourage researchers to adopt biomechanics approaches to design advanced physical medicine and rehabilitation approaches. Researchers should also consider using biomechanical approaches to evaluate the effectiveness of the interventions and to explore the mechanisms by which rehabilitation programs work for neuromusculoskeletal disorders. This Research Topic expanded the application of biomechanics, promoted the development of research on neuromusculoskeletal systems, to better understanding and treating neuromusculoskeletal disorders.

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Chronic ankle instability modifies proximal lower extremity biomechanics during sports maneuvers that may increase the risk of ACL injury: A systematic review

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The biomechanical changes in the lower extremity caused by chronic ankle instability (CAI) are not restricted to the ankle joint, but also affect the proximal joints, increasing the risk of joint injury. This study aimed to systematically review the research on CAI and lower extremity angle and movements during side-cutting, stop jumping, and landing tasks, to provide a systematic and basic theoretical basis for preventing lower extremity injury. Literature published from exception to April 2022 were searched in the PubMed, Web of Science, and SPORTDiscus databases using the keywords of "chronic ankle instability," "side-cut," "stop jump," and "landing." Only studies that compared participants with chronic ankle instability with healthy participants and assessed lower extremity kinetics or kinematics during side-cutting, stop jumping, or landing were included. The risk of bias assessment was conducted using a modified version of the Newcastle-Ottawa checklist. After title, abstract, and full text screening, 32 studies were included and the average score of the quality evaluation was 7 points (range 6–8). Among them five studies were related to the side-cut task, three studies were the stop-jump task, and twenty-four studies were related to landing. Although the results of many studies are inconsistent, participants with CAI exhibit altered lower extremity proximal joint movement strategies during side cut, stop jump, and landings, however, such alterations may increase the risk of anterior cruciate ligament injury.

KEYWORDS

ankle instability, lower extremity, joint biomechanics, sports activities, anterior cruciate ligament

1 Introduction

Ankle sprains are one of the most prevalent sports injuries, with 49.3% of ankle sprains occurring during sports (Fong et al., 2007). Ankle sprains, however, are frequently dismissed as minor ailments, with only around half of patients seeking professional help after their first sprain (Hubbard-Turner, 2019). Due to a lack of care and incorrect management, around 40%

of patients will develop to chronic ankle instability (CAI) after an initial ankle injury (Doherty et al., 2016). CAI is indicated by recurrent sprains, instances of ankle joint giving way, pain, swelling, and impaired function (Hertel and Corbett, 2019). The person with CAI has a higher risk of recurring sprains (Gribble et al., 2016). Approximately 68%–78% of individuals with CAI develop ankle osteoarthritis as a result of impaired balance and neuromuscular control of the lower extremity during movement (Moisan et al., 2017), which decreases the health-related quality of life and sports participation (Arnold et al., 2011).

Any impact on the ankle joint will affect the knee and hip joints since the lower extremities are a chain of movements. During ankle sprains, the mechanoreceptors in the joint capsule are damaged, and the impulse flow from the mechanoreceptors to the central nervous system is disrupted, causing problems with joint position and motion perception, as well as posture and gait reflexes (Wyke, 1967). As a result, a cascade of adaptive reactions is triggered, leading to alterations in movement patterns (Bullock-Saxton,1994). According to the lower extremity kinematic chain theory, the person with CAI demonstrates proximal joint kinematic modifications during sports, which are designed to compensate for the ankle joint's instability and functional impairment (Koshino et al., 2014).

Side cutting, stopping and landing are common sports movements, and they are also high-risk movements for anterior cruciate ligament (ACL) injury (Pflum et al., 2004; Chappell et al., 2007; Lin et al., 2009). Side cutting, which involves a change of direction and is a multi-plane movement, is common in court sports and also necessitates more complicated joint control for the lower limb than motions that occur in daily life (i.e. walking, running) (Koshino et al., 2016). While the stop jump task is a sport-related functional activity with a high risk of lower-limb injuries because the lower limb joints are required to endure the high impact energy exerted in a very short amount of time. The landing is typical of specific sports activity (Kim et al., 2018). Those sports maneuvers are more similar to real-life sports actions. Several studies comparing lower extremity biomechanical and neuromuscular control measures in CAI participants with healthy controls have been published, with some findings indicating that CAI participants had greater ankle dorsiflexion, knee flexion, and hip flexion (Caulfield and Garrett, 2002; Jeon et al., 2020), and prolonged peroneus longus latency during the landing phase of a single-leg drop (Simpson et al., 2019a). Terada et al. (2014a) demonstrated that participants with CAI demonstrated less knee flexion at peak anterior tibial shear force (ATSF) compared to the controls during stop jump. Altered lower extremity biomechanics due to CAI may make it much easier for patients to sustain non-contact ACL injuries (Theisen and Day, 2019). However, because our understanding of the complex pathological manifestations of CAI is still limited and many studies based on similar measurements have produced inconsistent results, there is still a necessity to better understand the mechanisms underlying recurrent sprains and limited sports participation in CAI. This study intends to use a systematic review to summarize, evaluate, and analyze the literature on sports biomechanics research on participants with CAI during side cuts, stops jump, and landing.

2 Methods

2.1 Search strategy

By April 2022, a literature search was conducted without regard to geographies, and publishing kinds. The databases searched included PubMed, Web of Science and SPORTDiscus. The computer search was supplemented with manual searches of the reference lists of all retrieved studies, review articles, and conference abstracts with the Related Articles tool. A Boolean logic method is used to write all database searches: (chronic ankle instability OR ankle instability OR functional ankle instability OR mechanical ankle instability) AND (lower limb OR lower extremity OR hip OR knee OR ankle) AND (kinematic OR kinetics OR biomechanics) AND: 1) side cut: (side cut OR side-cut OR side cutting OR side-cutting OR cut OR cutting), 2) stop jump: (stop jump OR stop-jump OR stop jumping OR stop-jumping), 3) landing: (land OR landing OR jump land OR jump landing OR jump-land OR jump-landing OR drop-vertical jump OR single-leg landing OR single-leg land OR jump OR jumping).

2.2 Inclusion and exclusion criteria

Two independent reviewers (BS and AHM) assessed titles and abstracts based on the inclusion criteria to identify eligible papers that would be subjected to full-text review. If the following criteria were met, studies were considered for full-text review: 1) published English language articles, 2) compared a CAI group to a healthy control group of participants, 3) main outcome measures were lower limb three joints angles and moments during side cut, stop jump, and landing, 4) study types were randomized controlled trials (RCTs) and retrospective comparative studies (including cohort or case-control studies), 5) if a study incorporated a therapy, it was only included if the control and CAI groups were compared before the intervention. Studies were excluded if the following criteria were met:1) non-English articles, 2) not compared with healthy controls, 3) main outcome measures were muscle activity, joint stiffness, energy dissipation, muscle energy production, ground reaction force, 4) animal experimental research, editorials, letters to the editor, review papers, case reports, and commentaries.

2.3 Risk of bias assessment

When disagreements arose, the same two independent authors (YX and GXN) assessed the study's quality and

discussed it together to reach a consensus. If the two authors were unable to reach an agreement, the final product was examined by a third author (CDZ). The Cochrane risk of bias tool was used to assess the methodological quality of RCTs (Higgins and Green, 2008). The modified Newcastle-Ottawa scale was used to assess the methodological quality of retrospective research (Taggart et al., 2001). The modified Newcastle-Ottawa scale has three components: patient selection, research group comparability, and outcome evaluation. Except for RCTs, each study was given a score between 0 and 9. Studies that received six or more points were deemed to be of good quality. The Centre for Evidence-Based Medicine in Oxford, United Kingdom, assessed studies based on the level of evidence they supplied (Howick et al. 2022).

2.4 Data extraction and analysis

Two of the authors independently gathered and summarized data from the included research (YX and GXN). The goal and quality of the study, participant characteristics, inclusion criteria, intervention protocols, and outcome variables were all assessed during the review process. The type of study, number and gender of participants, test site, test technique, foot condition, time frame, assessment parameters, assessment plane, lower extremity three-joint angle, and lower extremity three-joint moment were among the data retrieved. Any disagreement was resolved by the adjudicating senior authors (CDZ). No data pooling or meta-analysis was done since the included studies lacked homogeneity in terms of research methods, task site (flat or inverted), foot state (barefoot or shod), and CAI definition.

3 Result

3.1 Search results

A total of 70 articles related to side-cutting were initially searched in electronic databases, 53 articles remained after removal of duplicate articles, eight articles remained after the title and abstract review, and after full-text review, one article was excluded due to non-inclusion of healthy controls, and two articles were excluded due to no main observations given, and five articles were finally included. A total of 18 articles related to stop jump were initially searched, 14 articles remained after removal of duplicate articles, four articles remained after title and abstract review, and after full-text review, one article was excluded due to main outcome did not match, and three articles were finally included. A total of 246 articles related to landing were initially searched, 185 articles remained after removing duplicate articles, 29 articles remained after title and abstract review, and after full-text review, two articles were excluded due to not including healthy controls, and three articles were excluded due to not giving the main observation index, and 24 articles were finally included. For study selection, there was 90% agreement between the two authors. Figure 1 depicts a flow diagram of the study selection process, as recommended by PRISMA.

3.2 Characteristics of eligible studies

A total of 32 articles were included after searching for three different sports maneuvers, without randomized controlled studies, all of which were retrospective, and the characteristics of the included literature are shown in Table 1. In five publications, the test technique was side-cutting; all literature did a 45° side-cutting task, and shoe-wearing in two articles, the rest were not specified. The test method for the stop jump included three articles, all of which were tested with shoes on. The test procedure involves landing with 24 articles, a shoe-wear test of eight articles, and a barefoot test of six articles, the rest not stated. The platform height range was 15–70cm, with flat landing surfaces in 21 articles, slanted ground in four, and unstable terrain in one. Because certain studies did not give particular means, standard deviations, or mean differences in the text, this review does not provide specific data for this research.

3.3 Quality assessment

The mean score after scoring using the modified Newcastle-Ottawa scale (Taggart et al., 2001) was 7 (range 6–8, see Supplementary Table S1), with 97% agreement between the two reviewers.

3.4 Studies related to side cut

Only one of the five trials instructed participants to do a 45° sidestep-cutting maneuver following a straight approach run (Fuerst et al., 2018), whereas the other four asked them to do a forward jump and then a 45° crossover cut and run (Koshino et al., 2014; Koshino et al., 2016; Simpson et al., 2020b; Simpson et al., 2020a). Table 2 shows summaries of studies on side-cut. In the task which asked subjects to perform a forward jump and then a 45° crossover cut and ran, the CAI group exhibited significantly greater hip abduction (approximately 4.04°) from the 200 ms pre-IC (pre-initial contact) to 45% of the stance phase than the control group (Koshino et al., 2014), and greater hip flexion approximately 5.51° from 6% to 50% of the stance phase (Koshino et al., 2014), besides, Simpson et al. (2020a) found no differences between the groups in hip movements. In addition, the CAI group had considerably more knee flexion from 35 to 64% and 69%–87% of the stance phase than the control group,

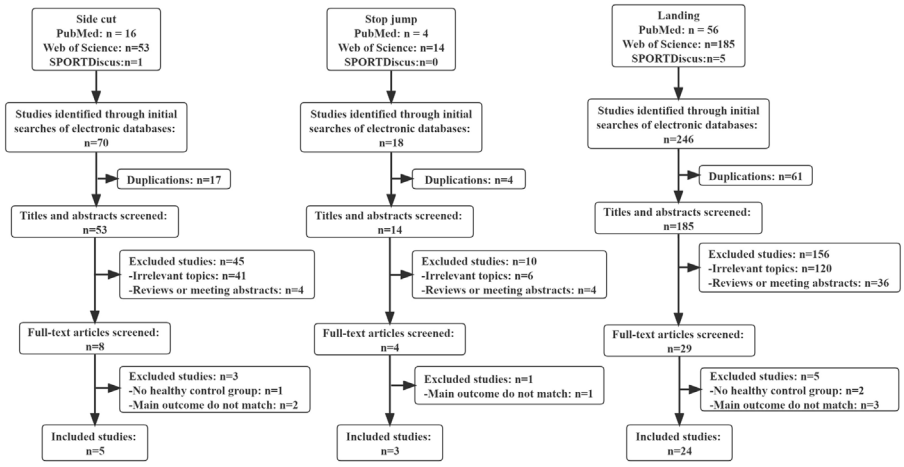

FIGURE 1
Flow chart of literature search and screening.

and the mean differences across groups were 7.63° and 9.54°, respectively (Koshino et al., 2014). In contrast, one study found no variations in knee angle between groups (Koshino et al., 2016). Only one study discovered that the CAI group had a lower knee abduction moment from 52% to 75% of the stance phase (difference = 0.27 ± 0.03 Nm/kg) (Simpson et al., 2020a). Almost the majority of the studies considered found differences in ankle angle or movement between the CAI group and healthy controls; only one research revealed no differences (Koshino et al., 2014). The CAI group exhibited significantly greater ankle inversion than the control group from 200 to 165 ms pre-IC (roughly 7.7°) and from 78 to 100% of the stance phase (roughly 6.4°) (Koshino et al., 2016). The CAI group had significantly greater ankle internal rotation from 35 to 54% of the stance phase compared to controls, with a mean difference of 6.62 ± 0.10° (Simpson et al., 2020b). The CAI group had a higher ankle plantar-flexion moment from 3% to 16% of the stance phase (difference = 0.22 ± 0.08 Nm/kg) and a lower ankle-eversion moment from 39% to 80% of the stance phase (difference = 0.13 ± 0.02 Nm/kg) than the control group (Simpson et al., 2020a). The findings of the task, which required individuals to complete a 45° sidestep-cutting maneuver after a straight approach run, only showed that the CAI group had smaller maximum inversion angles than the control group (Fuerst et al., 2018).

3.5 Studies related to stop jump

There were three studies in all, with two including ankle kinematics, two involving knee kinematics (Lin et al., 2011; Terada et al., 2014a), and just one involving hip kinematics (Terada et al., 2014a), however, no study involving lower extremity kinetics was included. Table 3 presents a summary of the key characteristics of the included studies. There was no difference in hip flexion at maximal anterior tibial shear force (ATSF) between the CAI and control groups (Terada et al., 2014a). When compared to the control group, the CAI group had a smaller knee flexion angle at IC, peak ATSF, and 100 ms post-IC, with mean group differences of 6.28°, 6.70°, and 7.30°, respectively (Terada et al., 2014a; Terada et al., 2014b). The CAI group exhibited a greater ankle inversion than the control group 140 ms post-IC. Furthermore, the CAI group's peak ankle eversion angle in the post-landing phase was lower than the control group's (difference = 3.40°) (Lin et al., 2011). At peak ATSF, however, there was no difference between the groups in ankle dorsiflexion (Terada et al., 2014a).

3.6 Studies related to landing

In total, twenty-four studies were included. All of them involved ankle kinematics and kinetics, 15 studies involving hip kinematics or kinetics (Delahunt et al., 2006; Gribble and Robinson, 2009; Gribble and Robinson, 2010; De Ridder et al., 2015; Son et al., 2017; Herb et al., 2018; Kim et al., 2018; Kunugi et al., 2018; McCann et al., 2019; Kim et al., 2019; Hopkins et al., 2019; Lin et al., 2019; Moisan et al., 2020; Jeon et al., 2020; Watanabe et al., 2022), and 17 studies involving knee kinematics or kinetics (Caulfield and Garrett, 2002; Delahunt et al., 2006; Gribble and Robinson, 2009; Gribble and Robinson, 2010; De Ridder et al., 2015; Son et al., 2017; Herb et al., 2018; Kim et al., 2018; Li et al., 2018; Kunugi et al., 2018; McCann et al., 2019; Kim et al., 2019; Hopkins et al., 2019; Lin et al., 2019; Moisan et al., 2020; Jeon et al., 2020; Watanabe et al., 2022). Supplementary Table S2 provides a summary of the characteristics of all included studies.

TABLE 1 Basic characteristics of the included studies.

Authors	Level of evidence	Design	Patients, no.		Matching[a]	Test method	Quality score
			CAI (M: F)	Control (M: F)			
Caulfield and Garrett, (2002)	—	NA	14 (14:0)	10 (10:0)	1,5	Single leg jump landing	7
Delahunt et al. (2006)	—	NA	24 (15:9)	24 (16:8)	3,4	Single leg jump landing	7
Gribble and Robinson, (2009)	4	CC	19 (10:9)	19 (10:9)	All	Double-leg take-off jump with a single-limb landing	8
Gribble and Robinson, (2010)	—	NA	19 (10:9)	19 (10:9)	All	Double-leg take-off jump with a single-limb landing	8
Lin et al. (2011)	4	CL	15 (6:9)	15 (7:8)	2,5	Vertical stop jump	7
Kipp and Palmieri-Smith, (2012)	—	NA	11 (5:6)	11 (5:6)	1,2,3,4,5	Double-leg take-off jump with a single-limb landing	6
Zhang et al. (2012)	—	NA	10	10	2,3,4	Double-leg take-off jump with a single-limb landing	7
Kipp and Palmieri-Smith, (2013)	—	NA	11 (5:6)	11 (5:6)	1,2,3,4,5	Double-leg take-off jump with a single-limb landing and 90°cut	6
Terada et al. (2014a)	4	CC	19 (10:9)	19 (10:9)	All	Vertical stop jump	8
Terada et al. (2014b)	4	CC	19 (10:9)	19 (10:9)	1,2,3,4,5	Vertical stop jump	7
Koshino et al. (2014)	4	CS	12 (10:2)	12 (10:2)	All	Forward jump and 45° crossover cut	8
Gehring et al. (2014)	—	NA	FAI + MAI: 19 FAI: 9	18	NA	Double-leg drop–vertical-jump	6
De Ridder et al. (2015)	—	NA	28 (10:18)	28 (10:18)	1,2,3,4,5	Double-leg take-off jump with a single-limb landing	7
Koshino et al. (2016)	4	CC	10 (9:1)	10 (9:1)	All	Forward jump and 45° crossover cut	8
Wright et al. (2016)	4	CS	23 (12:11)	23 (12:11)	1,2,3,4,6	Single leg jump landing	7
Son et al. (2017)	4	CL	22 (12:8)	22 (12:8)	1,3,4,5,6	Double-leg take-off jump with a single-limb landing and 90°cut	7
Fuerst et al. (2018)	4	CS	18 (8:10)	18 (8:10)	1,2,3,4	45°sidestep-cutting after a straight run	7
Herb et al. (2018)	4	CL	24	23	1,3,4,5,6	Double-leg drop–vertical-jump	7
Kim et al. (2018)	—	NA	100 (54:46)	100 (54:46)	All	Double-leg take-off jump with a single-limb landing and 90°cut	8
Li et al. (2018)	—	NA	21 (0:21)	21 (0:21)	All	Single leg landing	8
Kunugi et al. (2018)	4	CS	15 (15:0)	15 (15:0)	1,2,3,4	Double-leg take-off jump 45 °anterolateral with a single-limb landing and vertical-jump	7
McCann et al. (2019)	4	CC	25	25	2,3,4	Double-leg take-off jump with a single-limb landing	7
Kim et al. (2019)	—	NA	100 (54:46)	100 (54:46)	All	Double-leg take-off jump with a single-limb landing and 90°cut	8
Hopkins et al. (2019)	4	DL	200 (104:96)	100 (54:46)	5	Double-leg take-off jump with a single-limb landing and 90°cut	6
Lin et al. (2019)	—	NA	10 (8:2)	10 (8:2)	All	Single leg jump landing	8
Simpson et al. (2019b)	—	NA	15	15	1,3,4	Single leg landing	6
Moisan et al. (2020)	4	CC	32	31	1,2,3,4	Single leg jump landing	6
Jeon et al. (2020)	—	NA	MAI:10 (10:0) FIA:10 (10:0)	10 (10:0)	1,2,3,4,6	Double-leg take-off jump with a single-limb landing	6
Simpson et al. (2020a)	3	C	15 (7:8)	15 (7:8)	1,2,3,4,6	Forward jump and 45° crossover cut	6
Simpson et al. (2020b)	—	NA	15 (7:8)	15 (7:8)	1,2,3,4,6	Forward jump and 45° crossover cut	6
Watabe et al. (2021)	4	CS	12 (12:0)	12 (12:0)	1,2,3,4,6	Single leg landing	7
Watanabe et al. (2022)	—	NA	11 (7:4)	11 (7:4)	All	Single leg landing	8

[a]Comparability variables: 1 = gender; 2 = age; 3 = hight; 4 = weight; 5 = sports level; 6 = dominance side.

M, male; F, female; NA, data not available; C, Cohort Study; CC, Case-control study; CL, Control laboratory research; CS, Cross-sectional studies; DL, Descriptive laboratory studies.

TABLE 2 Summary of articles related to side cut.

Authors	Task	Foot condition	Planes	Time frame	Variables	Hip	Knee	Ankle
Koshino et al. (2014)	Forward jump and 45° crossover cut	Shod	F + S + T	(-) 200 m to	lower extremity joints angles	↑FL (6–50%SP) ↑ABD ((-) 200 m to 45%SP)	↑FL (35–64%, 69–87%SP)	NS
Koshino et al. (2016)	Forward jump and 45° crossover cut	Shod	F + S + T	(-) 200 m to	Lower extremity joints angles	↑FL (11–18%SP)	NS	↑INV((-)200 m to (-) 165 m, 78–100%SP)
Fuerst et al. (2018)	45°sidestep-cutting after a straight run	Shod	F	(-) 100 m, IC	Ankle joint angles and movements	—	—	↓peak INV
Simpson et al. (2020a)	Forward jump and 45° crossover cut	—	F + S + T	IC to	Lower extremity joints movements	NS	↓ABDM(52–75% SP)	↑PLM(3–16%SP) ↓EVM(39–81%SP)
Simpson et al. (2020b)	Forward jump and 45° crossover cut	—	F + S + T	IC to	Ankle joint angles	—	—	↑INT ROT (35–54%SP)

(-): before IC, (+): after IC, "-": not measured in the study.
ABD, abduction; ABDM, abduction movement; EVM, eversion movement; F, frontal; FL, flexion; IC, initial contact; INV, inversion; INT ROT, internal rotation; NS, no significant differences between groups; PFM, plantarflexion movement; S, sagittal; SP, stance phase; T, transversal; TO, teo off.

TABLE 3 Summary of articles related to stop jump.

Authors	Task	Foot condition	Planes	Time frame	Variables	Hip	Knee	Ankle
Lin et al. (2011)	Vertical stop jump	Shod	F	(-)200 m to (+) 200 m	Ankle angles	—	—	↑INV (0 m to (+) 140 m) ↓peak EV (0 to (+) 200 m)
Terada et al. (2014a)	Vertical stop jump	Shod	S	Peak ATSF	Lower extremity joints angles	NA	↓FL	NA
Terada et al. (2014b)	Vertical stop jump	Shod	S	(-)100 m, IC, (+) 100 m	Knee angels	—	↓FL (IC, (+) 100 m)	—

(-): before IC, (+): after IC, "-": not measured in the study.
ATSF, anterior tibial shear force; EV, eversion; F, frontal; FL, flexion; IC, initial contact; INV, inversion; NS, no significant differences between groups; S, sagittal.

The single-leg landing task was used in four studies (Li et al., 2018; Simpson et al., 2019b; Watabe et al., 2021; Watanabe et al., 2022). The mean platform height was 30 cm (range from 30 to 50 cm). As for the landing surfaces condition, two studies were flat (Watabe et al., 2021; Watanabe et al., 2022), one was a 20° inversion platform (Simpson et al., 2019b), and the other was a 5° inversion platform (Watabe et al., 2021). In studies using a flat landing surface (Watabe et al., 2021; Watanabe et al., 2022), no significant differences in hip angles or movements were identified between the CAI group and the control group (Watanabe et al., 2022). Peak knee flexion angle and peak ankle plantarflexion moment were bigger in the CAI group than in the control group in the 50 cm height platform condition, with a trend to be larger in the CAI group than in the control group in the 30 and 40 cm conditions. Furthermore, the CAI group had a greater peak ankle dorsiflexion angle than the control group at all heights. However, in the 30 cm height platform condition, Watanabe et al. (2022). (Watabe et al., 2021) showed no significant difference between groups for the greatest ankle inversion angle. There is no study on hip kinematics or kinetics when landing on an inverted surface; however, one study found that the CAI group had significantly higher knee values for flexion angle at IC (difference = 5.8°), peak flexion angle (difference = 15.7°), peak extension moment (difference = 0.27 Nm/kg), and peak internal rotation moment (difference = 0.12 Nm/kg) (Li et al., 2018). The CAI group had a 0.22 Nm/kg lower peak ankle eversion moment and a considerably larger maximum inversion angle than the control group (Li et al., 2018).

Five studies conducted single-leg jump landing tests with platform heights ranging from 35 to 46 cm (Caulfield and Garrett, 2002; Delahunt et al., 2006; Wright et al., 2016; Lin et al., 2019; Moisan et al., 2020). Only one study reported the distance from the platform to the center of the force plate was determined by the subject's leg length (Lin et al., 2019), others

were unclear. The landing surfaces in all five tests were flat, with only one including an unstable and 25° inversion platform (Moisan et al., 2020). When landing flat, compared to control, CAI patients exhibited a less externally rotated hip joint during a period from 200 to 55 ms pre-IC (Delahunt et al., 2006), as well as significantly larger hip flexion and hip adduction angle from initial contact to maximum knee flexion angle (Lin et al., 2019). Yet, Moisan et al. (Moisan et al., 2020) found no differences in hip angles across the groups. Furthermore, no difference in hip movements was identified across groups in any of the three studies. (Delahunt et al., 2006; Lin et al., 2019; Moisan et al., 2020). Caulfield et al. (Caulfield and Garrett, 2002) discovered that CAI had significantly more knee flexion than controls from 20 ms pre-IC to 60 ms post-IC, nevertheless, no difference was observed between groups for the remaining three studies (Delahunt et al., 2006; Lin et al., 2019; Moisan et al., 2020). Subjects with CAI demonstrated significantly greater ankle dorsiflexion from 10 ms pre-IC to 20 ms post-IC (Caulfield and Garrett, 2002), and less dorsiflexion from 90 to 200 ms post-IC (Delahunt et al., 2006). For the hindfoot, CAI displayed more dorsiflexion at IC (Wright et al., 2016). In the frontal plane, CAI individuals exhibited a more inverted ankle joint from 200 to 95 ms pre-IC (Delahunt et al., 2006), but a smaller ankle inversion angle and a significantly higher ankle eversion moment from initial contact to maximum knee flexion angle compared to controls (Lin et al., 2019). Moisan et al. (2020) on the other hand, found no differences in ankle joint angles and movement across groups. When landed on a laterally inclined surface, the CAI group showed greater knee extension moment during the landing phase compared to the control group and increased ankle dorsiflexion during the landing phase when landed on an unstable surface (Moisan et al., 2020).

Two studies performed a double-leg drop-vertical-jump task (DVJ) (Gehring et al., 2014; Herb et al., 2018). Participants in only one study were asked to perform a double-leg drop–vertical-jump task from a 30-cm box (Herb et al., 2018). In terms of hip angles and motions, there was no difference between CAI subjects and controls. From 95 to 200 ms post-IC, the CAI group had less knee flexion (difference = 8.23 ± 0.97°), however, there were no differences in knee joint moments between groups. The CAI group had more ankle inversion from 107 to 200 ms post-IC (difference = 4.01 ± 2.55°) and less plantar flexion from 11 to 71 ms post-IC (difference = 5.33° ± 2.02°). From 11 to 77 ms post-IC (difference = 0.17 ± 0.09 Nm/kg) and from 107 to 200 ms post-IC (difference = 0.23 ± 0.03 Nm/kg), the CAI group had a higher plantar-flexion moment (Herb et al., 2018). Another study asked participants to land at the platform with 24° inversion and 15° plantar flexion. They discovered that the maximum ankle inversion was much higher in CAI (difference = 5.5°) compared to the control group (Gehring et al., 2014).

Seven studies involved a double-leg take-off jump followed by a single-limb landing (Gribble and Robinson, 2009; Gribble and Robinson, 2010; Zhang et al., 2012; Kipp and Palmieri-Smith, 2012; De Ridder et al., 2015; McCann et al., 2019; Jeon et al., 2020). The platform height varied from 15 to 70 cm, and the distance between the platform and the force plate's center was variable. All of the studies' landing surfaces were flat. Two of them discovered that the CAI group had significantly greater hip joint flexion angles during landing than the controls (Jeon et al., 2020), and less hip abduction from 78 ms pre-IC to 34 ms post-IC (McCann et al., 2019). By contrast, the other three studies found no differences between groups in hip angles (Gribble and Robinson, 2009; Gribble and Robinson, 2010; De Ridder et al., 2015) or internal joint moments for hip joints between groups (Jeon et al., 2020). When compared to the CAI group, the control group produced significantly more knee flexion 100 ms pre-IC (difference = 94.29°) (Gribble and Robinson, 2010). The control group showed more knee flexion than the CAI group at IC (difference = 4.42°) (Gribble and Robinson, 2009). During the landing phase, the CAI group had more knee flexion than the control group (Jeon et al., 2020). Unlike, the other three studies found no significant difference in knee joint angles and movements between groups (De Ridder et al., 2015; McCann et al., 2019; Jeon et al., 2020). Impressively, none of the seven studies found any significant differences in ankle joint angles and moments across groups (Gribble and Robinson, 2009; Gribble and Robinson, 2010; Kipp and Palmieri-Smith, 2012; Zhang et al., 2012; De Ridder et al., 2015; McCann et al., 2019; Jeon et al., 2020).

Five studies conducted double-leg take-off jump with a single-limb landing and 90°cut task (Kipp and Palmieri-Smith, 2013; Son et al., 2017; Kim et al., 2018; Kim et al., 2019; Hopkins et al., 2019). In each of the five studies, the landing surface was flat. Ground contact was divided into two parts during this task: the landing phase, which lasted from first ground contact to peak joint flexion, approximately 0%–50% of stance, and the side-cutting phase, which lasted from the end of the landing through takeoff, approximately 51%–100% of stance. In the case of the hip joint, relative to controls, CAI patients decency with more hip flexion angle (Son et al., 2017; Kim et al., 2019; Hopkins et al., 2019), more hip abduction (Son et al., 2017; Kim et al., 2019; Hopkins et al., 2019), higher hip extension moment (Son et al., 2017; Kim et al., 2018; Hopkins et al., 2019), and less hip abduction moment (Son et al., 2017; Hopkins et al., 2019) (Figure 2). In terms of the knee joint, CAI patients displayed a tendency with more knee flexion (Son et al., 2017; Kim et al., 2019; Hopkins et al., 2019), more knee abduction angle (Son et al., 2017; Hopkins et al., 2019), less knee extension moment (Son et al., 2017; Kim et al., 2018; Hopkins et al., 2019), and less knee abduction (Son et al., 2017; Hopkins et al., 2019) moments relative to controls (Figure 3). As for ankle joint, results are inconsistent across studies, Figure 4 summarizes results from each study.

Only a single study performed a double-leg take-off jump 45° anterolateral from a 30 cm height platform with a single-limb

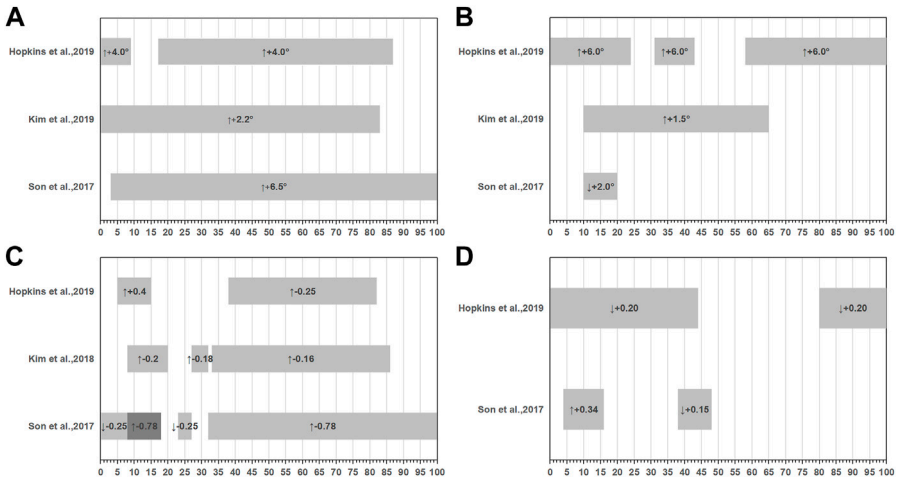

FIGURE 2
Summary of hip joint kinematics or kinetics during double-leg take-off jump with a single-limb landing and 90° cut task. **(A)** Sagittal hip angles, **(B)** Frontal hip angles, **(C)** Sagittal hip movements, **(D)** Frontal hip movements. (-): extension, (+): flexion or abduction, (↑): variables increased compared to the health group, (↓): variables decreased compared to the health group.

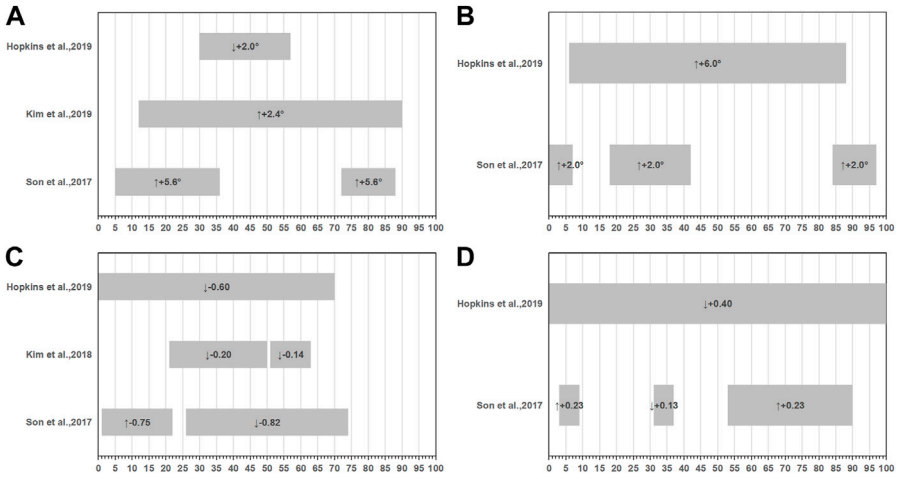

FIGURE 3
Summary of knee joint kinematics or kinetics during double-leg take-off jump with a single-limb landing and 90° cut task. **(A)** Sagittal knee angles, **(B)** Frontal knee angles, **(C)** Sagittal knee movements, **(D)** Frontal knee movements. (-): extension, (+): flexion or abduction, (↑): variables increased compared to the health group, (↓): variables decreased compared to the health group.

landing at the flat and vertical-jump task. When compared to the control group, the CAI group had significantly less hip adduction (difference = 3.99°) from 70 to 120 ms post-IC, less knee flexion (difference = 7.33°) from 120 to 190 ms post-IC, less knee external rotation (difference = 4.14°) from 300 to 250 ms pre-IC, and less ankle dorsiflexion (difference = 12.83°) from 10 ms pre-IC to 300 ms post-IC (Kunugi et al., 2018).

4 DISCUSSION

This study was designed to systematically review and evaluate the effects of CAI on lower extremity joint angles and moments during side-cut, stop jump, and landing tasks. This study found that the biomechanical changes of the proximal lower extremity joints caused by CAI may increase the risk of ACL injury during

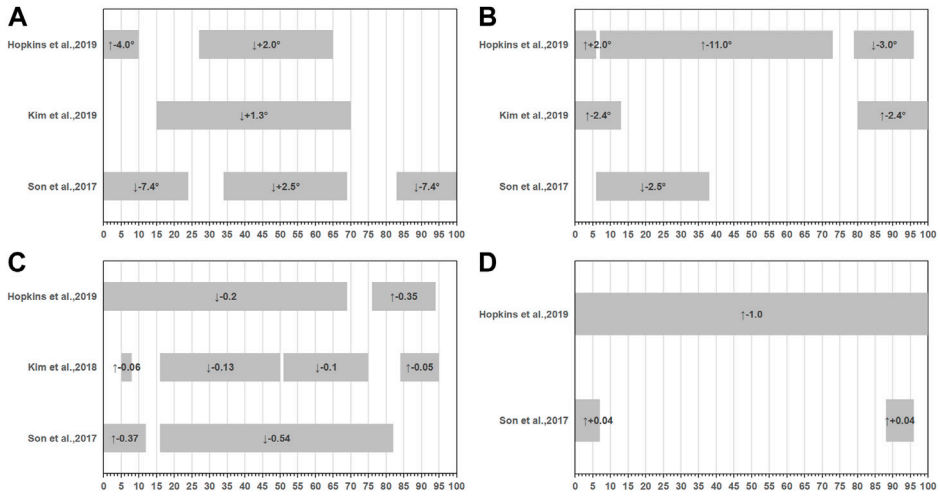

FIGURE 4
Summary of ankle joint kinematics or kinetics during double-leg take-off jump with a single-limb landing and 90° cut task. **(A)** Sagittal ankle angles, **(B)** Frontal ankle angles, **(C)** Sagittal ankle movements, **(D)** Frontal ankle movements. (-): plantarflexion or inversion, (+): dorsiflexion or eversion, (↑): variables increased compared to the health group, (↓): variables decreased compared to the health group.

high-risk movements of ACL injury, such as side-cut, stop jumps, and landing.

4.1 Changes in ankle biomechanics

The greater trend observed in the ankle dorsiflexion position in the CAI subjects during the early phase after IC is a trend towards greater protection of the lateral ligamentous complex because in the dorsiflexion position, the ligaments is unlikely to be stretched. During the landing procedure, subjects may have been subconsciously striving to reduce the consequences of ground impact forces on their ligaments. In a recent systematic review, it was hypothesized that increased ankle dorsiflexion during jump landing for people with CAI could be due to centrally mediated motor program changes caused by the unstable ankle, which would place the talocrural joint in a tightly packed position to protect the lateral ankle ligaments from excessive inversion (Simpson et al., 2019b).

However, CAI subjects were not as efficient as control group subjects in obtaining the closed-packed dorsiflexed posture of the ankle joint during the late phase after IC. This could indicate a restriction in the posterior talar glide. Limited dorsiflexion range of motion has been identified as a factor in the development of lateral ankle joint injuries (Tabrizi et al., 2000). The decrease in peroneal longus activity before IC leaves the ankle joint in a vulnerable position (i.e., a more inverted position) and induces a hyper inversion injury in CAI participants (Delahunt et al., 2006). Since the ankle is in a less dorsiflexion position it is unable to absorb forces as well and greater forces are transmitted to the knee, increasing the risk factors that can lead to non-contact knee injuries (Theisen and Day, 2019).

CAI patients had a more inverted foot position after IC, which was attributed to a lack of ankle proprioception (Lin et al., 2011) and diminished peroneus muscle activity (Delahunt et al., 2006). Because of subtalar instability, this higher inversion ankle position at IC has also been linked to an increased risk of ankle roll-over and sprain injury (Yamamoto et al., 1998). The greater the ankle inversion angle, the greater the risk of recurring lateral ankle sprain (Lin et al., 2011). When the lateral ankle is loaded during IC, reduced ankle eversion moments indicate that the lateral ankle muscle is unable to control frontal plane movement eccentrically (Son et al., 2017; Kim et al., 2018). This could also cause excessive inversion of the ankle complex, resulting in a recurrent lateral ankle sprain.

4.2 Changes in knee biomechanics

Participants with CAI had a small flexion angle of the knee joint (Terada et al., 2014b) after IC. A straightened knee has less potential energy decay and could indicate a lack of absorption capabilities (Herb et al., 2018). However, there may be a compensatory strategy that allows CAI participants to absorb kinetic energy from their bodies. When the ankle and knee joints land on the ground, doing work on the eccentric part of the sagittal frontal plane will help to reduce the body's kinetic energy in the vertical direction (Norcross et al., 2013). Increased knee extension at impact may provide the ground reaction forces more time to dissipate and control. These findings of changes in knee

biomechanics following ground impact suggest that the presence of CAI may alter the distal-to-proximal connection that provides an efficient and effective system for transmitting forces up the kinetic chain (Dejong et al., 2020). Decreased knee flexion angle can result in insufficient energy attenuation capabilities of the knee, resulting in the knee joint receiving large compressive impact forces, increasing the stress on the ATSF and the load on the ACL (Chappell et al., 2005; Norcross et al., 2010). Nevertheless, those biomechanical changes in the knee joint are related to the mechanism of ACL injury (Fleming et al., 2001) and may increase the risk of ACL injury.

4.3 Changes in hip biomechanics

Due to induced sensorimotor loss at the ankle, CAI participants demonstrate increased hip joint dependency to maintain balance and stability during those three sports maneuvers (Horak et al., 1990). Subjects with CAI may attempt to adjust to a position relative to their low center of mass for dynamic stability, primarily utilizing hip flexion (Koshino et al., 2014). CAI may be able to obtain enough balance to adequately stabilize themselves in the sagittal plane by increasing hip flexion motion to mitigate the effects of GRF (Lin et al., 2019). Increased hip flexion angles can help the eccentrically controlled hip extensors absorb or dissipate the impact more effectively over time. Participants with CAI may attempt to land safely with more flexed hip positions, protecting the unstable ankle from the high-impact landing (Son et al., 2017). It may be easier and safer for CAI patients to have their femurs more vertically upright (less abducted), which may help stabilize the downward motion of their center of body mass in the sagittal plane. A compensatory load redistribution strategy from the unstable distal (e.g., ankle) to proximal (e.g., hip) joints, indicates that the hip joint in the sagittal plane may play an important role in sports maneuvers (Son et al., 2017). However, CAI patients displayed higher hip adduction over the majority of the task's stance phase, possibly due to decreased frontal plane hip joint stability (Son et al., 2017). Increased hip abduction may prevent excessive ankle inversion motion (Koshino et al., 2014). People with CAI may have developed a strategy to alter frontal plane hip kinematics to compensate for ankle instability during the task (Dejong et al., 2020).

4.4 Reason for inconsistent results

Although this systematic review discovered that CAI participants demonstrate a changed movement strategy during sports actions involving cutting, stopping jumping, and landing, these changes are not limited to the ankle joint, but rather manifest farther up the kinematic chain (e.g., knee, and hip). The clinical variability among studies was found to be significant in this systematic review. Disparities of this nature might be classified into five categories. The first category is concerned with the various authors' definitions of CAI. Different inclusion criteria were applied in different studies, resulting in a non-homogeneous population. As a result, comparing data from different studies is challenging. The International Ankle Consortium established selection criteria guidelines to follow when researchers assess participants with CAI in 2013 (Gribble et al., 2013). However, it should be noted that several of the studies included in this systematic review (8/32) were published before these principles were widely recognized. The second category refers to the sports level of participants. Half of the included studies (15/32) had recreational physical activity participants as subjects, while others (6/32) had athletes as subjects. However, some research classified recreational exercise as "at least 20 min of strenuous activity, three or more days per week" (Terada et al., 2014a; Terada et al., 2014b) while others were "at least 30 min of exercise three times per week." (Gribble and Robinson, 2009). The third category refers to discrepancies in the experimental tasks, such as the height of the platform and the surface landing condition. The fourth category refers to the food conditions in which the participants performed the experimental tasks. Some studies measured kinematics and kinetics data while the subjects were wearing shoes, while others measured them while they were barefoot. The fifth category is related to various data processing methods. Lower limb kinematics or kinetics parameters were calculated at initial contact or point of maximum vertical ground reaction force in some studies, while some studies used the one-dimensional statistical parametric mapping (SPM) analysis. The external validity of the outcomes of this systematic review is compromised by all of these variations. As a result, there was no way to pool the data and perform a meta-analysis.

5 Conclusion

The changed angle and movements of the knee and hip found in CAI participants during sports maneuvers involving side cutting, stop jumping, and landing might be a result of central nervous system modifications following a peripheral ankle joint injury (Ward et al., 2015). Because of the mechanical advantages of the proximal joints, participants with CAI may evolve an altered movement strategy to redistribute the force from the unstable distal joint (e.g., ankle) to the more stable proximal joints (e.g., knee and hip). This shift in movement pattern, on the other hand, transmitting greater forces to the knee, which increases the load on ACL, may be linked to an increased risk of ACL injuries.

Data availability statement

The original contributions presented in the study are included in the article/Supplementary Material, further inquiries can be directed to the corresponding authors.

Author contributions

YX, BS, and AM wrote the manuscript. BS and AM assessed titles and abstracts based on the inclusion criteria. YX and GN assessed the risk of bias and collected the data. CZ resolved any disagreement. BS made critical revisions related to the important intellectual content of the manuscript.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Supplementary material

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fphys.2022.1036267/full#supplementary-material

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Eight-week virtual reality training improves lower extremity muscle strength but not balance in adolescents with intellectual disability: A randomized controlled trial

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Purpose: This study aims to assess the effects of 8-week virtual reality (VR) training on balance and lower extremity muscle strength in adolescents with intellectual disability (ID).

Methods: Thirty adolescents with intellectual disability were randomly divided into the virtual reality group and control group. The participants in the virtual reality group and the control group received the virtual reality training and the physical education (PE) course, respectively, for 8 weeks. The Berg Balance Scale (BBS), Timed Up and Go (TUG) test and lower extremity muscle strength were measured before and after the training.

Results: The between-group results showed that the participants in the virtual reality group increased the muscle strength of hip flexors ($p < 0.001$), hip extensors ($p = 0.002$), hip abductors ($p < 0.001$), knee flexors ($p < 0.001$), knee extensors ($p = 0.002$) and ankle plantar flexors ($p = 0.042$) significantly after training, compared to the control group. However, no significant improvement was found in the berg balance scale and timed up and go scores between the virtual reality group and control group after training ($p > 0.05$). The within-group results showed that the strength of all the muscle groups significantly increased after training in the virtual reality group ($p < 0.05$) compared to the baseline. However, no significant difference was found in the muscle strength in the control group before and after training. The within-group berg balance scale and timed up and go scores showed no significant improvements in both groups.

Conclusion: Virtual reality training intervention might be effective in improving the lower extremity muscle strength, but no significant improvement was found on balance ability in adolescents with intellectual disability.

KEYWORDS

virtual reality, intellectual disability, balance, muscle strength, adolescents

1 Introduction

Intellectual disability (ID) is a developmental disorder characterized by limitations on intellectual functioning and adaptive behaviour. These limitations arise before the age of 22 (Schalock et al., 2021). The prevalence of ID is approximately 1% (McKenzie et al., 2016). This disability may affect the nervous and/or sensory system and cause functional and physical disability (Manel Font-Farré et al., 2021). Previous studies showed that high level of physical fitness during adolescence could be a predictor of high level of physical activity in adulthood (Pertti Huotari et al., 2011). Some studies suggest that individuals with ID have a low level of physical activities compared with their typically developing peers. This might indicate that the physical fitness in ID adolescents is relatively low since they have a low level of physical activities. In addition, low level of physical fitness, including poor muscular strength and balance may result in impaired stability and increase risk of falls (Westendorp et al., 2011; Blomqvist et al., 2013; Einarsson et al., 2015; Hocking et al., 2016; Hsu., 2016; Wouters et al., 2020). Individuals with ID are prone to falls and injuries (Enkelaar et al., 2012). A review paper reported that the proportion of people with ID who were subjected to falls was 39% [95% CI (0.35–0.43)](Ho et al., 2019). For example, individuals with ID are more likely to have fall-related fractures due to low bone mineral density. Compared with the healthy population, the rate of hospitalization due to injuries was as twice as high in the ID population, and most of these injuries were fall-related (Enkelaar et al., 2012). Poor balance control might be related to the falls in the ID population. (Hsu, 2016). Balance control is closely related to lower extremity muscle strength (J C Nitz et al., 2010). Good muscle strength and balance control are crucial for preventing falls in physical activities of daily life, such as walking and stair climbing (Muehlbauer et al., 2015; Obrusnikova et al., 2021). Therefore, improving muscle strength and balance control is necessary for individuals with ID.

Different trainings can have positive effects on motor function; however, they also have limitations. Several physical therapies have been used to improve muscle strength and balance in individuals with ID, including resistance training (Cowley et al., 2011; Obrusnikova et al., 2021), sensorimotor gymnastics (Jankowicz-Szymanska et al., 2012), balance training (Lee et al., 2016), combined training program (Kachouri et al., 2016), physical education (PE) (Gupta et al., 2011) and others. Obrusnikova et al. and Cowley reported that resistance training could significantly increase the muscular strength of individuals with ID (Cowley et al., 2011; Obrusnikova et al., 2021). Jankowicz-Szymanska found that systematic sensorimotor gymnastics can improve static balance ability in young people with ID (Jankowicz-Szymanska et al., 2012). Lee et al. found that balance training had the effects on improving the postural balance in adolescents with ID (Lee et al., 2016). In addition, Kachouri et al. suggested a training program combined strength and proprioceptive training had the effects on improving the muscle strength and postural balance in children with ID (Kachouri et al., 2016). However, the review paper (Jeng et al., 2015) showed that exercise training is ineffective in improving balance and coordination in adolescents with ID. The effects of physical training on balance and muscle strength are inconsistent. In addition, PE is an important opportunity to promote exercise in adolescents with ID. PE is widely used as a control condition in previous studies, which aimed to investigate the effects of specific training programs in improving the muscle functions and balance (Gupta et al., 2011). For example, Gupta et al. found that regular physical activities followed at school may improve the strength and balance in children with ID (Gupta et al., 2011). And another study also showed PE can improve lower extremity muscle strength in students with ID (Hsu., 2016). However, evidence for isolated or limited social interactions and less physical engagement during PE course were found in students with ID compared to their healthy peers (Erica Gobbi et al., 2018). Less social interactions and physical engagements may be attributed to the lack of enjoyment for adolescents with ID during PE course. Consequently, an attractive and effective physical training is necessary to improve the balance and muscle strength for individuals with ID.

Virtual reality (VR) plays an increasingly important role in motor rehabilitation (Juras et al., 2019). VR is defined as "the use of interactive simulations created by computer to provide users with environments" (Weiss et al., 2006). The use of interactive video games as a form of motor rehabilitation can improve motivation, exercise performance, and tolerance in children (Ferguson et al., 2013). It stimulates the brain in a multisensory manner by creating temporary interaction scenarios that involve all the senses, and adds motivation and pleasure in a game-like environment, which creates good training experiences (Leeb and Perez-Marcos, 2020). The convergence of VR gaming with exercise have shown good results in physical, mental and social health (Nathan Keller et al., 2022). Therefore, VR training may have better training compliance compared to traditional physical trainings methods. Previous study showed that a minimum of 8 weeks of VR training should be done for the training effect to show up (Lee et al., 2019). Few studies showed the effect of VR training on balance and lower extremity muscle strength in adolescents with ID. In previous studies, VR has been used to improve the functional abilities of individuals with ID (Patrice L Tamar Weiss et al., 2003; Shira Yalon-Chamovitz and Weiss, 2008; Lotan et al., 2009; Ahn, 2021). Shira et al. showed that VR appeared to offer diverse and stimulating experiences during leisure activities among individuals with ID. In addition, VR can be easily combined with physical training, which is highly adaptable and suitable for the ID population (Shira Yalon-Chamovitz and Weiss, 2008). Lotan et al. found that VR training had the effects on improving the physical fitness of

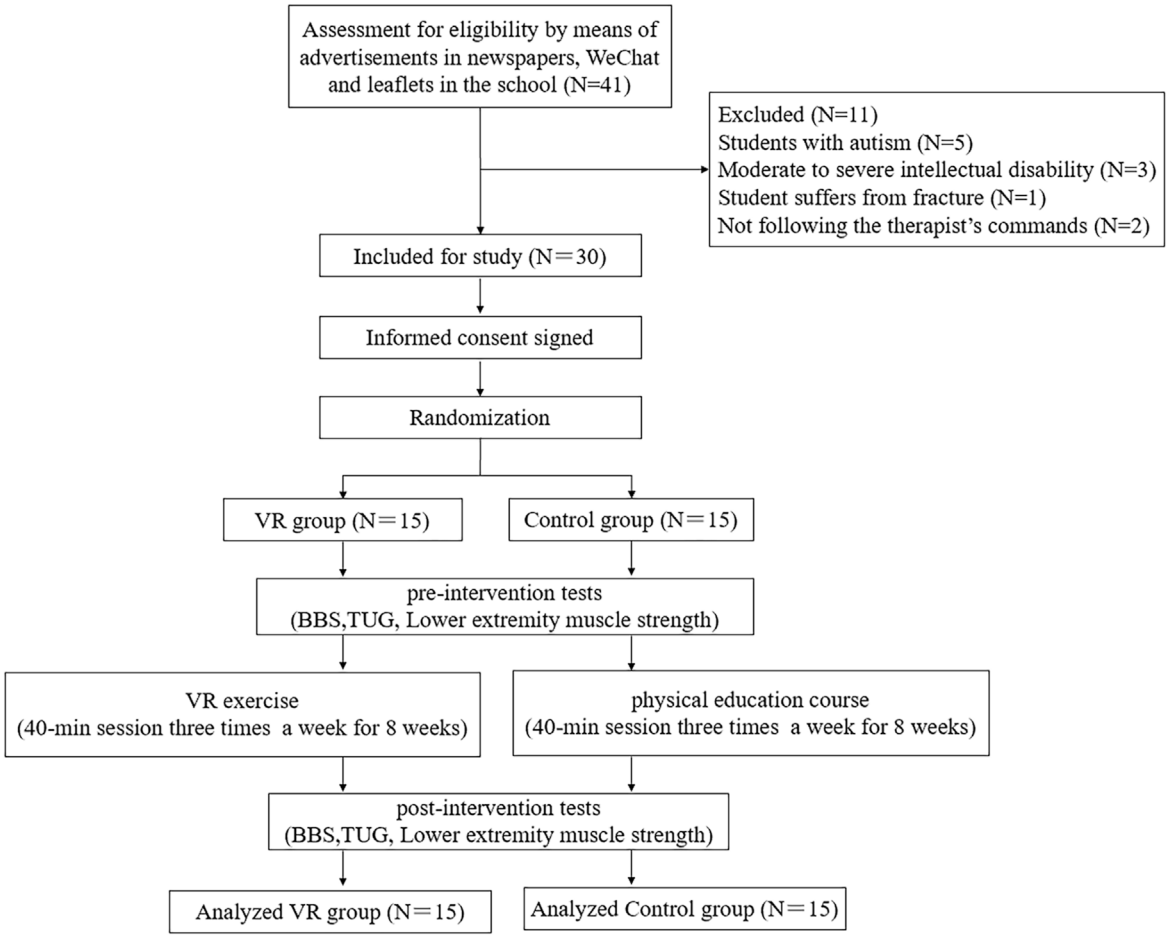

FIGURE 1
Flow diagram for randomized controlled trial.

individuals with ID (Lotan et al., 2009). Ahn showed that VR and computer game-based cognitive therapy for visual-motor integration is an effective training method for children with ID to promote visual perception and motor function (Ahn, 2021). Furthermore, some studies have shown that VR training can improve the balance and muscle strength of older adults and individuals with neurological diseases, such as patients suffering from stroke, Parkinson's disease and cerebral palsy (Cho et al., 2016; Lee et al., 2019; Mak and Wong-Yu, 2019; Orsatti-Sánchez. and Diaz-Hernandez., 2021; Sadeghi et al., 2021).

The present study aimed to investigate the effects of VR training on balance and lower extremity muscle strength in adolescents with ID. We hypothesized that b: 1) The participants in VR group and PE group would have significant improvement on balance and lower extremity muscle strength after training. 2) The VR training could be more efficient than PE course for improving balance and lower extremity muscle strength in adolescents with ID.

2 Materials and methods

2.1 Study design

A randomized controlled trial was designed to evaluate the effect of 8-week VR training on balance and lower extremity muscle strength (Figure 1). The participants in VR group and control group received the VR training and PE course for 8 weeks, respectively. Muscle strength and balance control were measured at the baseline and immediately after training.

2.2 Participants

2.2.1 Sample size estimation

G*Power software (Version 3.1.9.2) was used to estimate the sample size. The following data were determined: effect size = 0.57 (Hsu., 2016), two-tailed significance, statistical power = 0.8,

FIGURE 2
The virtual reality training system.

α = 0.05. Accounting for a drop-out rate of 20% (Wang et al., 2021), 15 participants were recruited per group in this study.

2.2.2 Participant recruitment and randomization

In this study, 41 adolescents were contacted from a special education school. A total of 30 participants (male = 11, female = 19) aged 11–18 years old who were diagnosed with ID by means of advertisements in newspapers, WeChat and leaflets in the school met the study eligibility criteria and participated in this study. All the participants were randomly assigned to the VR group (n = 15) and control group (n = 15). The Wechsler Intelligence Scale was used to measure the intelligence quotient (IQ) of the participants (Jamile Benite Palma Lopes et al., 2017). The inclusion criteria (Eid, 2015; Lee et al., 2016; Ahn, 2021) were: 1) adolescents whose IQ ranged from 50 to 70; 2) adolescents who can follow the therapist's commands; and 3) adolescents who can independently stand and walk. The exclusion criteria (Cowley et al., 2011; Lee et al., 2016; Ahn, 2021) were: 1) any contraindications to exercise; 2) severe intellectual, visual and hearing disability; and 3) any musculoskeletal, neurological, cardiovascular or respiratory system disorders. This study was approved by the ethics committee of Shandong Sport University. All the parents were requested to sign a written informed consent statement before the study started. The study was registered in the Chinese Clinical Trial Registry (ChiCTR1900021352).

2.3 Exercise intervention

The frequency and duration of VR training were designed to be consistent with the educational program outline of the school. The participants in VR group received the 40-min VR training session 3 times a week for 8 weeks. Each session included a 5-min warm-up, 30-min intervention training and 5-min cooldown. The warm-up included 5-min of walking and stretching. The intervention training included the VR training. The cooldown included the stretching exercises. The participants were required to attend at least 24 sessions for the VR training.

The participants in the VR group were trained with a VR training system (Xbox Kinect, Shanghai), which included Kinect, console, dance carpet, white screen and projector (Figure 2). The Xbox Kinect, console and monitor were installed in a quiet room. The participants were instructed to remain standing at a distance of 1.5–3.5 m in front of the VR training system. Before the training started, the therapist showed the participants how to use the system. In each session, all the participants played five sports games that the therapist selected in advance among 15 sports according to enjoyment level. The five video games included cool-running, whack a mole, catching gift, racing and playing football, which were designed to enhance the muscle strength, endurance, balance and coordination of adolescents with ID. Each game lasted for 6 min. The participants were individually taught to play the VR training games by the therapist in the first six sessions. After the participants learned how to play the training games, they independently practiced VR training games in the rest of the sessions.

The participants in the control group received the 40-min PE course session 3 times a week for 8 weeks. The PE course included 5-min of warm-up, 30-min of exercise (playing activities, jogging and jumping rope) (Gupta et al., 2011), and 5-min of cooldown. The first six PE course were leaded by a PE teacher who assisted and guided the participants during activities and ensured safety. After that, the participants exercised independently in the PE course during the rest of the sessions.

2.4 Outcomes

2.4.1 Balance control

The Berg Balance Scale (BBS) and Timed Up and Go (TUG) test were the most common clinical tests to measure the balance control. These measurements were conducted in a quiet testing room (Juras et al., 2019).

BBS is one of the most commonly used scales in the clinical practice (Sibley. et al., 2011). It is a reliable and standardized tool for balance. It is reported that BBS reliable with the inter-rater reliability estimated at 0.97 (95% CI 0.96–0.98) and intra-rater reliability estimated at 0.98 (95% CI 0.97–0.99) (Downs, 2015). BBS included 14 task items: reaching forward with an outstretched arm, standing with eyes closed with one foot in front, turning, retrieving an object from the floor, standing on one leg, sitting to stand, turning 360°, standing, placing the alternate foot on a stool, transferring, standing with feet together and standing to sitting unsupported (Ekiz et al., 2015). The rating score for each task ranged from 0 to 4 (0 =

TABLE 1 Muscle strength testing protocol.

Muscle group	Participant position	Dynamometer position
Hip flexors	Sitting, hip flexed at 30°off surface	Resistance given anteriorly to distal thigh
Hip extensors	Prone, knee and thigh extended off surface, pelvis stabilized	Resistance given to posterior distal thigh
Hip abductors	Lateral, hip extented at 20°	Distal fibula
Knee flexors	Sitting, knee flexed at 45°	Heel
Knee extensors	Sitting, knee flexed at 90°	Anterior proximal to lateral malleolus
Ankle plantarflexors	Supine, knee flexed at 90°	Across metatarsal heads

unable to perform the task; 4 = task is performed independently) (Muir-Hunter et al., 2015), with a maximum score of 56. A greater score in BBS indicated better balance (Downs, 2015).

TUG is a modified test for functional walking. It is a simple method to investigate the dynamic balance. It had a high test-retest reliability (ICC = 0.932) (Martin et al., 2017). The participants stood up from a chair, walked a length of 3 m, turned, walked back and sat on the chair. The time between when the researcher said "go" and when the participant's back touched the backrest of the chair was recorded in seconds. The total time was recorded with a stopwatch. The test was repeated three times, and the fastest time was recorded. A less time in completing the TUG test indicated better dynamic balance control.

2.4.2 Lower extremity muscle strength

The dominant lower extremity was defined as the preferred leg for kicking a football (Gribble. et al., 2007). The muscle strength of hip flexors, hip extensors, hip abductors, knee flexors, knee extensors and ankle plantar flexors of the dominant leg was measured using a hand-held dynamometer (HHD) (MicroFET3, United States) (Gupta et al., 2011; Wuang et al., 2013) (Table 1). The unit of muscle strength was Newton (N). The ICC values of HHD were 0.95 (hip flexors), 0.84 (hip extensors), 0.96 (knee flexors), 0.93 (knee extensors) and 0.69 (ankle plantar flexors) (Wuang et al., 2013). The researcher held the HDD stationary while the participant applied a maximum force against the dynamometer. Participants were asked to perform test with the maximum force 2–3 s and to maintain until the researcher said to relax. The participants had 30 s of rest between the consecutive trials. Three successful trials were collected for each subject. The collected data were averaged for data analysis.

2.5 Statistical analysis

The data of participants who attended all three evaluation sessions were included for the final data analysis. All the data were presented using mean ± standard deviation. All data were tested for normality using the Shapiro-Wilk test. Two-way analysis of variance (ANOVA) with repeated-measures was used to compare the outcome data between groups before and after training. If an interaction effect was found, the follow-up tests with Bonferroni adjustments were performed. SPSS 25.0 (SPSS Inc., IL, United States) was used for statistical analysis. A p-value less than 0.05 indicated significant difference.

TABLE 2 The baseline characteristics of the participants.

	VR group (n = 15)	Control group (n = 15)	p
N (male/female)	15 (5/10)	15 (6/9)	0.705
Age (years)	13 ± 0.8	14 ± 2.2	0.087
Height (cm)	1.67 ± 0.11	1.60 ± 0.10	0.069
Weight (kg)	65.46 ± 20.90	61.73 ± 10.26	0.541

Note: VR = virtual reality.

3 Results

3.1 Baseline characteristics of the participants

A total of 30 participants (15 in the VR group, and 15 in the control group) completed the entire 8-week study without loss of participants. No significant differences were found between the two groups in age, height and body weight (Table 2).

3.2 Balance control

Two-way ANOVA with repeated-measures showed no significant interaction effect in BBS ($p > 0.05$) and TUG test ($p > 0.05$) between groups before and after the training (Table 3).

3.3 Lower extremity muscle strength

Two-way ANOVA with repeated-measures showed a significant interaction effect in the muscle strength of hip extensors ($p = 0.043$), hip abductors ($p = 0.015$) and knee flexors ($p = 0.006$) (Table 4). The follow-up tests showed that muscle strength was significantly

TABLE 3 The balance ability in adolescents with ID (X ± S).

	VR group		Control group		*p*		
	Pre-test	Post-test	Pre-test	Post-test	Group	Time	Time × group
BBS (score)	50.66 ± 3.56	52.07 ± 2.94	50.12 ± 4.25	50.62 ± 3.21	0.269	0.269	0.604
TUG (seconds)	9.15 ± 2.01	9.08 ± 2.09	9.42 ± 1.15	9.73 ± 2.01	0.336	0.811	0.700

Note: BBS = berg balance scale; TUG = timed up and go; VR = virtual reality.

TABLE 4 The results of the lower limb muscle strength in adolescents with ID (Newton).

	VR group		Control group		*p*		
	Pre-test	Post-test	Pre-test	Post-test	Group	Time	Time × group
Hip flexors	15.85 ± 1.70	17.00 ± 2.10*△	14.29 ± 2.13	14.27 ± 1.87	<0.001	0.258	0.249
Hip extensors	12.03 ± 1.85	13.87 ± 1.21*△	11.80 ± 2.11	11.68 ± 2.03	0.075	0.013	0.043
Hip abductors	11.99 ± 2.11	14.43 ± 1.72*△	11.58 ± 1.80	11.57 ± 1.83	0.010	0.015	0.015
Knee extensors	14.06 ± 1.51	16.01 ± 1.83*△	13.05 ± 1.55	13.70 ± 1.44	<0.001	0.020	0.114
Knee flexors	14.05 ± 1.66	17.08 ± 1.90*△	12.63 ± 1.36	13.39 ± 1.22	<0.001	<0.001	0.006
Ankle plantarflexors	12.06 ± 1.72	14.01 ± 1.25*△	11.79 ± 1.73	12.81 ± 1.50	0.076	0.010	0.259

Note: VR = virtual reality. *Denotes a significant difference compared with the pre-test value within each group, $p < 0.05$. △Denotes a significant difference between the VR, and control group, $p < 0.05$.

improved by 7.3% for hip flexors ($p < 0.001$), 15.3% for hip extensors ($p = 0.008$), 20.4% for hip abductors ($p = 0.001$), 21.6% for knee flexors ($p = 0.001$), 13.9% for knee extensors ($p < 0.001$) and 16.2% for ankle plantar flexors ($p = 0.001$) in the VR group after training compared to the baseline. No significant change in muscle strength was found in the control group after training compared to the baseline ($p > 0.05$). Compared with the control group, significant improvements in the muscle strength were found in hip flexors ($p < 0.001$), hip extensors ($p = 0.002$), hip abductors ($p < 0.001$), knee flexors ($p < 0.001$), knee extensors ($p = 0.002$) and ankle plantar flexors ($p = 0.042$) after the training.

No significant time×group interaction effects were found on the muscle strength of hip flexors ($p = 0.249$), knee extensors ($p = 0.114$) and ankle plantar flexors ($p = 0.259$). Moreover, the main effect of group was found for hip flexors and knee extensors. The analysis of group main effect showed that the muscle strength of hip flexors ($p < 0.001$) and knee extensors ($p < 0.001$) were significantly greater in the VR group than in the control group. After the training, the significant main effect of time was observed on the knee extensors ($p = 0.02$) and ankle plantar flexors ($p = 0.01$).

4 Discussion

The aim of this study was to investigate the effect of 8-week VR training on balance and lower extremity muscle strength in adolescents with ID. The results of the study partially supported the two hypotheses.

The results of present study showed that lower limb muscle strength significantly improved after the 8-week VR training in VR group. However, no significant improvement was found in the control group after training. These results partially support the first hypothesis. This finding was consistent with the results of a previous study, in which lower limb muscle strength improved in children with cerebral palsy compared with the control group after an 8-week VR training program (Cho et al., 2016). Another study also reported regular VR training improved the physical fitness in individuals with ID (Lotan et al., 2009). In addition, Carlos et al. reported that individuals with ID improved performance in the virtual task (Carlos Bandeira de Mello Monteiro et al., 2017). Jamile et al. reported the transcranial direct current stimulation combined with Xbox-Kinect game improved the upper limb movement in individuals with ID (Jamile Benite Palma Lopes et al., 2020). Such findings can be explained by the following reasons. First, VR training involves movements with muscle contractions, which may have the strengthening effects. Second, VR training enhances high-intensity, repetitive and task-oriented training (Lee et al., 2019). Third, VR training exhibits other advantages compared to PE course. For example, VR training has been used in neurological rehabilitation as a novel therapeutic method that combines physical activity with the environment. An enriched environment can stimulate the acquisition of motor skills and

partially repair neuronal impairment in people with ID (De Giorgio, 2017). This may also be related to visual stimulation. The visual cortex has become the most widely used system for studying the mechanisms of brain plasticity in adulthood (WIESEL. and HUBEL., 1963). VR training provides continuous sound and visual stimulation during practice, acting as a direct support, which in the long term facilitates motor learning (Silas de Oliveira Damasceno et al., 2022). A previous study (Ahn, 2021) found that VR and computer game-based therapy are effective in enhancing the visual perceptual and motor functions. In this study, participants were provided with visual stimulation by seeing images projected onto a screen, which may enhance the strengthening effect. Fourth, improvements in function induced by VR games are thought to be related to interactive training with a wide range of activities and scenarios involving multiple sensory channels and the creation of exercises (Jamile Benite Palma Lopes et al., 2022). And the interaction provided by VR game devices serves as an interesting, encouraging and safe environment.

It is noteworthy that the balance control of the two groups was not significantly improved after training in this study. These findings did not support the first hypothesis. Similar results were found in previous studies that training did not have the effect on improving the balance control in individuals with cerebral palsy or Parkinson's disease (Wu et al., 2019; Chen et al., 2020). Another study also showed no significant improvement was found in the TUG test score in the VR group and PE group after 8 weeks of training (Hsu, 2016). None of the exercises in the VR training included the activities with spinning movements or maintaining balance with eyes closed. These activities may challenge the balance control of the participants to improve the balance. Lack of these activities may explain that no significant difference was found in BBS and TUG test after training. In the future, these activities should be included in the VR training programs to improve the balance control of individuals with ID. TUG test is a comprehensive test to assess the functional mobility, which requires muscle strength, proprioception, executive function, attention, coordination and control (Mirelman et al., 2018). In one previous study, TUG was significantly improved after 9-week of VR training in the VR group compared to the control group (Yousefi Babadi and Daneshmandi, 2021). In addition, another study also demonstrated significant improvement in TUG after VR training (Steven Phu et al., 2019). This contradiction to our study may be due to the difference in the duration of intervention, as well as the difference in the participants. Thus, our further study should consider to prolong the training periods for the effect of balance improvement to show up. Lower limb muscle strength is a crucial factor for maintaining static and dynamic balance (Lee et al., 2016). However, the present study found that the significantly improvement in lower limb muscle strength but not balance ability in VR group. Therefore, the improvement in muscle strength may not directly related to the balance improvement. Other factors, such as involvement of central nervous system in balance contorl, should be considered in future study.

In this study, no adverse events were reported during the training sessions. Thus, the training program should be safe for adolescents with ID. All the participants completed all the training sessions without drop-outs, which indicated that this type of training had good compliance for adolescents with ID. People with ID are typically inactive, and thus, their physical functions, such as endurance, balance and strength, are less trained than those of their peers without ID (Enkelaar et al., 2012). The enhancements of muscle strength, balance and physical fitness are socially important for adolescents with ID (Shields et al., 2008).

This study had several limitations. The sample size in this study was small, which limited the generalization of the findings. Thus, future study with a larger sample is necessary. In this study, individuals with Down syndrome were included in the sample and could affect the results because they present specific characteristics in muscle force and postural control. In addition, the reliability of HHD measuring muscle strength can be affected by the position of the test (Lue. et al., 2009). Therefore, we should be careful when measuring muscle strength with HDD to ensure the position of the test. Finally, a more objective measure of balance should be included in future study.

5 Conclusion

According to the results of this study, we concluded that the 8-week VR training intervention can improve lower extremity muscle strength, but not balance of adolescents with ID. Adolescents with ID in special education school are recommended to add regular VR training in addition to participating in PE course to improve muscle strength. Further studies should be suggested to prolong intervention duration for improving balance.

Data availability statement

The raw data supporting the conclusion of this article will be made available by the authors, without undue reservation.

Ethics statement

The studies involving human participants were reviewed and approved by the ethics committee of Shandong Sport University. Written informed consent to participate in this study was provided by the participant's; legal guardian/next of kin.

Author contributions

SW provided research design, data collection, data analysis, data interpretation and writing. HY provided data acquisition, data processing, and data interpretation. ZL provided research design, data analysis and paper revision. JW provided research design, data analysis, data interpretation, paper revision, final approval of the version to be submitted. All authors read, critically reviewed, and approved the final version of the manuscript.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Running economy and lower extremity stiffness in endurance runners: A systematic review and meta-analysis

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Background: Lower extremity stiffness simulates the response of the lower extremity to landing in running. However, its relationship with running economy (RE) remains unclear. This study aims to explore the relationship between lower extremity stiffness and RE.

Methods: This study utilized articles from the Web of Science, PubMed, and Scopus discussing the relationships between RE and indicators of lower extremity stiffness, namely vertical stiffness, leg stiffness, and joint stiffness. Methodological quality was assessed using the Joanna Australian Centre for Evidence-Based Care (JBI). Pearson correlation coefficients were utilized to summarize effect sizes, and meta-regression analysis was used to assess the extent of this association between speed and participant level.

Result: In total, thirteen studies involving 272 runners met the inclusion criteria and were included in this review. The quality of the thirteen studies ranged from moderate to high. The meta-analysis results showed a negative correlation between vertical stiffness (r = −0.520, 95% CI, −0.635 to −0.384, $p < 0.001$) and leg stiffness (r = −0.568, 95% CI, −0.723 to −0.357, $p < 0.001$) and RE. Additional, there was a small negative correlation between knee stiffness and RE (r = −0.290, 95% CI, −0.508 to −0.037, $p = 0.025$). Meta-regression results showed that the extent to which leg stiffness was negatively correlated with RE was influenced by speed (coefficient = −0.409, $p = 0.020$, $r^2 = 0.79$) and participant maximal oxygen uptake (coefficient = −0.068, $p = 0.010$, $r^2 = 0.92$).

Conclusion: The results of this study suggest that vertical, leg and knee stiffness were negatively correlated with RE. In addition, maximum oxygen uptake and speed will determine whether the runner can take full advantage of leg stiffness to minimize energy expenditure.

KEYWORDS

spring, vertical stiffness, leg stiffness, joint stiffness, running economy

Introduction

Running is a popular sport around the world. Statistics from the US government show that more than 55.9 million people in 2017 were involved in running, jogging, and trail running (Statista, 2018). Khammassi et al. (2020) state that these regular, high-intensity physical activities offer many health benefits. In another study, Ezzatvar et al. (2022) stated that running can reduce the risk of infection and improve immune function, especially during pandemics such as the COVID-19 pandemic. Although running is considered an effective approach to maintain the health of the body, runners use it to enhance their athletic performance. There are many complex factors affecting the performance of runners. These factors include maximum oxygen uptake, lactate threshold and running economy (Joyner, 1991). Running economy (RE) seems to be the most appropriate parameter to distinguish between the endurance running performance of untrained and trained runners (Conley and Krahenbuhl, 1980; Williams and Cavanagh, 1987; Santos-Concejero et al., 2014). Running economy is the oxygen uptake at a given submaximal running speed (Conley and Krahenbuhl, 1980; Morgan and Craib, 1992; Anderson, 1996). Some studies define RE as the energy cost per unit distance (Rodrigo-Carranza et al., 2021). Running economy is one of the most important factors in determining endurance running performance (di Prampero et al., 1986).

When a person is running, bones, muscles, tendons, ligaments and other elements are usually modeled as Spring Mass Model (SMM) to support the body running forward (Blickhan, 1989; Butler et al., 2003). When a person is running, sufficient lower extremity stiffness is required to maintain motor performance. The mechanical stiffness of the lower extremity may be reflected by the vertical, leg, and joint stiffness (McMahon and Cheng, 1990; Butler et al., 2003).

Vertical stiffness describes the vertical displacement of the center of mass (COM) in response to vertical ground reaction force during a task performed in the sagittal plane (Latash and Zatsiorsky, 1993). The vertical stiffness is considered the first stiffness parameter to be measured, and the models for leg and joint stiffness were expanded (Morin et al., 2005; Brughelli and Cronin, 2008). The measurement of vertical stiffness requires the least amount of equipment in the experiment and can be obtained fast (Maloney and Fletcher, 2021). However, the calculation of vertical stiffness only attempts to model the cumulative stiffness of the lower extremity holistically and does not consider the effects of specific details. Leg stiffness characterizes the structural components of the leg, including muscles, tendons, and ligaments (Kerdok et al., 2002), which are an important component of the stretch-shortening cycle (SSC). These structural components of the leg reflect the compression of the leg spring in any plane or direction corresponding to the force (McMahon and Cheng, 1990; Hunter and Smith, 2007). During running, the leg touches the ground at a certain angle (θ). Therefore, the leg spring compression is greater than the COM displacement. Some studies have found that the leg stiffness is always less than the vertical stiffness (Farley and González, 1996; Matt; Brughelli and Cronin, 2008). Also, when the theta angle is 0 during vertical jumping, it leads to a leg stiffness that produces the same value as the vertical stiffness (Beerse and Wu, 2017). Many researchers refer to the results calculated using the equation for vertical stiffness as leg stiffness, which may cause confusion among readers about this concept (Maloney and Fletcher, 2021; Struzik et al., 2021).

The vertical stiffness and leg stiffness were calculated using the whole lower extremity as a whole SMM (Blickhan, 1989; Butler et al., 2003), and such measurements do not take into account multiple degrees of freedom of the lower extremity. Therefore, Farley and his colleagues proposed a joint stiffness calculated with the torsion spring model. First, the researchers deconstructed the hip, knee and ankle joints into three torsion springs (Farley et al., 1998). Then, the relative contribution of the stiffness of the three joints to the overall leg spring stiffness was evaluated (Farley et al., 1998). Joint stiffness describes the resistance to changes in angular displacement in flexion and rotation after applying a joint moment (Latash and Zatsiorsky, 1993; Butler et al., 2003; Brughelli and Cronin, 2008). The inverse dynamics principle obtains this net joint moment. By using a torsional spring model, joint stiffness values can be estimated for the hip, knee and ankle joints during vertical and horizontal motion. Previously, neither the vertical nor the leg stiffness could detect how each joint relative contributes to the stiffness of the whole leg. Joint stiffness in a running task can be used to consider the relative contribution of each joint to total stiffness (Butler et al., 2003; Maloney and Fletcher, 2021).

Researchers started exploring the relationship between stiffness and energy expenditure in the 1990s. However, there is still no higher quality evidence for the relationship between lower extremity stiffness and energy expenditure, and some studies have even found opposite results for each other (Tam et al., 2018; Zhang et al., 2022). Furthermore, no scholar has yet reviewed specifically the relationship between lower extremity stiffness and RE. If lower extremity stiffness is related to RE, these results may be a key indicator to evaluate the exercise performance of endurance runners. The purpose of this study was to explore whether lower extremity stiffness was associated with RE in endurance runners.

Methods

This review followed the Preferred Reporting Items for Systematic Review and Meta-Analyses (PRISMA) statement for reporting systematic reviews.

Search strategy

The literature search was conducted on three databases, PubMed, Web of Science, and Scopus. Studies published

between 1980 and 2022 were screened. Journal article titles, abstracts, and keywords in each database were searched using the following terms and Boolean operators: (running economy OR energy cost OR metabolic cost OR energetics OR VO_2 OR VO_{2max} OR cost of running OR consumed oxygen OR oxygen uptake) AND (leg OR lower limb OR lower extremity OR vertical OR joint) AND (stiffness).

The study was conducted in two stages. First, two independent reviewers evaluated and included potential studies based on titles and abstracts. Second, the selected research papers were categorized as meeting the inclusion criteria (yes), likely to be included (maybe) and not meeting the inclusion criteria (no). Finally, disagreements between the two independent reviewers were discussed and resolved in a consensus session. A third reviewer was consulted where consensus could not be reached.

Inclusion criteria and data extraction

The articles included in this study met the following criteria: 1) Healthy middle-distance runners aged >18 years with a maximal oxygen uptake (VO_{2max}) > 50 ml/kg/min. 2) The study included the observation of both RE and lower extremity stiffness metrics. 3) Correlations between lower-extremity stiffness and RE, such as Pearson's correlation coefficient or r-square, are reported.

On the other hand, papers were excluded if they 1) were conference presentations, posters, or case studies; 2) included non-runners (e.g., athletes in basketball, and soccer); 3) did not provide data or inferred data on RE or biomechanical characteristics; and 4) did not report relevance.

After the full-text screening, a custom table was created to record basic information about the study (e.g., age, country, height, VO_{2max}, and test speed). In addition, Pearson's correlation coefficient (r) was extracted for the relevant stiffness parameter and energy expenditure. For studies reporting r-square, the square root of the coefficient of determination was taken by its directionality and was transformed into a correlation coefficient.

Quality evaluation

The methodological quality of each study was assessed using the Joanna Briggs Institute's (JBI) Analytical Cross-Sectional Study Critical Assessment Tool (Moola et al., 2017). The checklist consists of ten items to evaluate the suitability of studies for inclusion in systematic reviews and meta-analyses. Two assessors assigned each study a score of 2 for "yes," 0 for "no," and 1 for "unclear." Studies with a total score greater than 70% were considered to be of high quality.

Data analysis

A comprehensive meta-analysis version 3.0 (Englewood, NJ, United States) was used to analyze pooled Pearson's correlation coefficients. A p-value of less than 0.05 was considered significantly different. The correlation coefficient was used to summarize results and report 95% confidence intervals. The heterogeneity of studies was assessed using the I^2 statistic. Studies with I^2 of less than 50% had poor heterogeneity, and a fixed-effects model was considered; otherwise, a random-effects model was used (Buccheri et al., 2018). According to Hopkins evaluation scheme, correlation size was judged as follows: <0.1 = trivial, 0.1–0.29 = small, 0.30–0.49 = moderate, 0.50–0.69 = large, 0.70–0.89 = very large, and ≥0.90 = extremely large (Hopkins et al., 2009). A simple meta-regression analysis was performed on the two independent variables, VO_{2max} and test speed, to determine the variation of the moderating variables on the magnitude of the correlation. The correlation coefficients for each study were converted to Fisher's Z-score as the dependent variable for the regression analysis. To assess the effect of individual studies on the pooled correlation coefficients and to test the robustness of the correlation between running economy and lower extremity stiffness, sensitivity analyses were performed by sequentially reducing one study per round. Publication bias was examined using funnel plots, and Duval and Tweedie's trim and fill correction were used to address the impact of publication bias on the main meta-analysis.

Result

Study identification and characteristics

In total, 1,608 articles were obtained by searching three databases including PubMed, Web of Science, and Scopus. The Endnote software was used to remove duplicates, and 553 articles remained. In addition, papers with only titles and abstracts were filtered, and 27 articles with full texts remained. These articles were further screened. Out of the 27 articles, one longitudinal article included an intervention category, five did not include energy consumption or stiffness metrics, and eight did not include an available correlation coefficient. These articles were excluded. As a result, thirteen articles were included in our systematic review (Dalleau et al., 1998; Heise and Martin, 1998; Slawinski et al., 2008; Rabita et al., 2011; Barnes et al., 2014; Lazzer et al., 2014; Man et al., 2016; Rogers et al., 2017; Tam et al., 2018; Monte et al., 2020; Hansen et al., 2021; Li et al., 2021; Zhang et al., 2022), as shown in Figure 1.

In total, 272 runners were included in this study, 90.44% male and 9.56% female, with a mean age of 24.93 ± 6.48 years. The VO_{2max} of the participants included in the study was 54.02–71.5 ml/kg/min. Ten studies reported correlations between vertical stiffness and RE, ten other studies reported

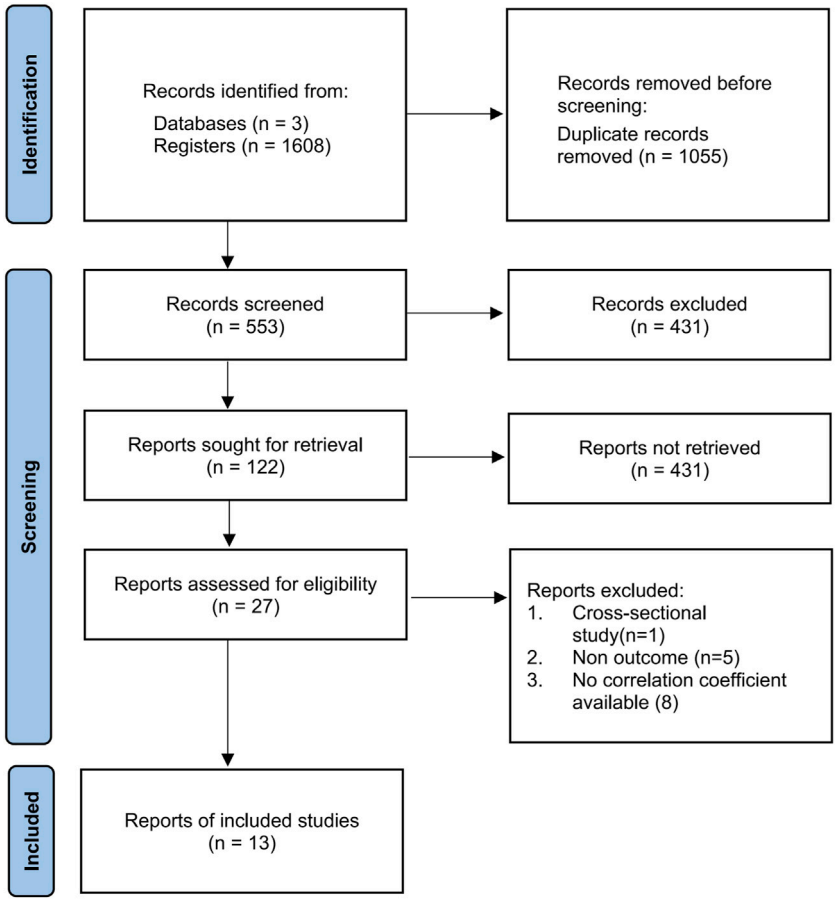

FIGURE 1
PRISMA flow diagram showing the selection process of eligible articles.

correlations between leg stiffness and RE, and two studies reported correlations between knee stiffness and ankle stiffness and RE.

To acquire vertical, leg, and joint stiffness, the majority (n = 11) of the studies were conducted during running. Barnes and his colleagues used the peak force measured during five squat jumps divided by the vertical displacement as the "leg stiffness" (Barnes et al., 2014), which we summarized as the vertical stiffness. The approach used for measuring vertical stiffness was adopted by Rogers et al. (2017). In all studies, leg stiffness and joint stiffness were measured during running. Details of the included studies are shown in Table 1.

Quality evaluation

Table 2 shows the quality evaluation scores for each study (Table 2). The mean JBI quality rating score for the thirteen studies included in the analysis was 15.85 ± 0.86. The majority of these studies (n = 12) were of high quality. One study had a low rating because it lacked ethical consideration. A common low-scoring item in the entries was "How was the study population selected?" Given that the purpose of this review was to evaluate a simple correlation between lower extremity stiffness and RE, the sampling and selection of the study population did not affect the results of this study. In particular, most studies had clear objectives, valid and reliable data collection, and appropriate statistical methods. Therefore, no study was excluded because of methodological quality issues.

Pooled analyses

Ten studies reported Pearson correlation coefficients between vertical stiffness and RE. Heterogeneity between vertical stiffness and RE across studies was small (I^2 = 20.62%). As a result, a meta-analysis was performed using a

TABLE 1 Characteristics of the included studies.

Author	Country	Participant (male)	Age (years)	VO_{2max} (ml/kg/min	Intensity (m/s)	Unit of measure (RE)
Zhang et al. (2022)	China	30 (30)	20.00–22.00	54.02 ± 4.67	2.78	ml/kg/min
Li et al. (2021)	China	28 (28)	20.70 ± 1.20	65.78 ± 4.99	3.33	ml/kg/min
					3.89	
					4.44	
Hansen et al. (2021)	Denmark	12 (12)	22.40 ± 3.10	67.04 ± 4.2	3.89	ml/kg/min
					5.00	
Monte et al. (2020)	Italy	32 (32)	37.90 ± 13.00	Endurance runners	3.56 ± 0.34	J/kg/m
Tam et al. (2019)	South Africa	30 (30)	258.00 ± 5.00	Trained runners	3.30	ml/kg/min
Rogers et al. (2017)	Australia	11 (11)	20.00 ± 2.90	67.60 ± 3.80	3.89	Kcal/kg/km
Man et al. (2016)	China	9 (9)	20.00 ± 3.90	Experienced runners	2.87	ml/min $kg^{-0.75}$
Lazzer et al. (2014)	Italy	15 (15)	40.50 ± 8.40	55.20 ± 6.70	2.78	ml/kg/m
Barnes et al. (2014)	New	39 (39)	20.80 ± 2.80	68.70 ± 4.80	3.89	ml/kg/min
	Zealand	24 (0)	20.50 ± 2.10	59.90 ± 3.50		
Rabita et al. (2011)	France	9 (6)	23.20 ± 3.20	71.50 ± 6.50	5.10 ± 0.30	J/kg/m
Slawinski et al. (2008)	France	9 (7)	NR	67.80 ± 5.20 (male) < 59.30 ± 4.70 (female)	5.25 ± 0.40	ml/kg/m
Heise et al. (1998)	United States	16 (16)	27.30 ± 4.80	62.20 ± 3.00	3.35	ml/kg/min
Dalleau et al. (1998)	France	8 (8)	24.00 ± 5.00	65.60 ± 4.60	5.10 ± 0.33	ml/kg/m

fixed effects model. The results revealed a large pooled correlation between vertical stiffness and RE (r = -0.519, 95% CI, −0.619 to −0.402, $p < 0.001$), as shown in Figure 2. Ten studies reported Pearson correlation coefficients between leg stiffness and RE. The test for heterogeneity between leg stiffness and RE was moderate (I^2 = 67.93%). As a result, a random-effects model was adopted for meta-analysis. Pooled results showed a large pooled correlation between leg stiffness and RE (r = −0.568, $p < 0.001$, 95% CI, −0.723 to −0.357), as shown in Figure 3.

Two studies reported pooled results for knee stiffness and ankle stiffness. Pooled results with small heterogeneity showed a small negative correlation between knee stiffness and RE (r = −0.290, 95% CI, −0.508 to −0.037, p = 0.025). In contrast, ankle stiffness was not associated with RE (r = 0.0838, 95% CI, −0.716 to 0.788, p = 0.86), and interestingly, the two studies presented opposite results. As shown in Figure 4.

Further one-way meta-regression analyses were performed for studies to provide clear reporting of test velocity and participant VO_{2max}. The results showed that the correlation between vertical stiffness and RE was not affected by participant motion VO_{2max} (coefficient = 0.003, p = 0.854) and test velocity (coefficient = 0.046, p = 0.751). However, the negative correlation between leg stiffness and RE increased with the VO_{2max} (coefficient = −0.068, 95% CI, −0.114 to −0.022, p = 0.010, r^2 = 0.92) test velocity (coefficient = −0.409, 95% CI, −0.730 to −0.089, p = 0.020, r^2 = 0.79), as shown in Figure 5.

Sensitivity analyses and publication bias

We performed a sensitivity analysis with one-by-one exclusion and the estimated effects would still be within the 95% CI of the pooled results. Funnel plots of effect sizes versus standard errors were generated to determine possible publication bias. Analysis of the funnel plot revealed that the included studies were generally located on the left side of the pooled results. The funnel plot on the leg stiffness and RE correlations was adjusted using Duval and Tweedie's trim and fill correction to produce a symmetric funnel plot around the Pearson correlation coefficient. This correction shifted the overall effect size to the right but did not change the main results, which still showed a significant trend, as shown in Figure 6.

Discussion

This study is the first to examine the relationship between lower extremity stiffness and RE systematically. The aim was to investigate whether lower extremity stiffness affects RE. In total, 13 articles were included in this review. The results showed a significant negative correlation between vertical, and knee stiffness and RE in endurance runners. This result suggests that vertical, leg, and knee stiffness influence RE in endurance runners. In addition, the results of the meta-regression showed that the correlation coefficient between leg stiffness and RE was related to the runner's speed and VO_{2max}.

TABLE 2 Methodological quality evaluation for the included studies.

Author	Item 1	Item 2	Item 3	Item 4	Item 5	Item 6	Item 7	Item 8	Item 9	Item 10	Total
Zhang et al. (2022)	2	0	0	2	2	2	2	2	2	1	15
Li et al. (2021)	2	0	0	2	2	2	2	2	2	2	16
Hansen et al. (2021)	2	0	0	2	2	2	2	2	2	1	15
Monte et al. (2020)	2	0	0	2	2	2	2	2	2	2	16
Tam et al. (2017)	2	0	0	2	2	2	2	2	2	2	16
Rogers et al. (2017)	2	0	0	2	2	2	2	2	1	2	15
Man et al. (2016)	2	1	1	1	2	2	2	2	2	2	17
Lazzer et al. (2014)	2	0	2	2	2	2	2	2	2	1	17
Barnes et al. (2014)	1	1	0	2	2	2	2	2	2	2	16
Rabita et al. (2011)	2	1	0	2	2	2	2	2	2	1	16
Slawinski et al. (2008)	2	0	0	2	2	2	2	2	2	2	16
Heise et al. (1998)	2	1	0	2	2	2	0	2	2	1	14
Dalleau et al. (1998)	2	1	0	2	2	2	2	2	2	2	17

Item 1: Is the research purpose of the study clear? Are the arguments sufficient?
Item 2: How was the study population selected? (whether the research subjects were randomly selected and whether stratified sampling was adopted to improve the representativeness of the sample).
Item 3: Are the inclusion and exclusion criteria for the sample clearly described?
Item 4: Are the sample characteristics clearly described?
Item 5: Are the data collection tools reliable and valid? (If an investigator survey is taken, how reproducible is the survey result?)
Item 6: What are the measures to verify the authenticity of the data?
Item 7: Are ethical issues considered?
Item 8: Is the statistical method correct?
Item 9: Is the presentation of the findings appropriate? (Are the results and inferences distinguishable, and are the results faithful to the data rather than inferences).
Item 10: Is the research value clearly articulated?

Vertical stiffness and running economy

Current findings showed that vertical stiffness negatively correlated with RE during sub-maximal running. The following formula was used to calculate the vertical stiffness:

$$k_{vert} = \frac{F_{max}}{\Delta y}$$

Where F_{max} is the peak vertical force, and Δy is the vertical displacement of the COM from ground contact until mid-stance.

Many studies have demonstrated that increasing the peak vertical force or decreasing the COM vertical displacement during running will result in better vertical stiffness. Several studies have found that increasing running speed leads to a higher peak vertical force and a decrease in COM vertical displacement (Matt Brughelli and Cronin, 2008; Brughelli et al., 2011; Halvorsen et al., 2012). Dutto and Sam (2002) found that when the runner is exhausted, the change in vertical stiffness is primarily related to the maximum body displacement in the vertical direction during running rather than the change in peak vertical force. Other studies have also found that runners reduce energy expenditure by exhibiting less COM vertical displacement (Cavanagh et al., 1977; Anderson, 1996; Heise and Martin, 2001; Halvorsen et al., 2012). They observed that the peak vertical force decreases while the vertical stiffness increases significantly with fatigue (Morin et al., 2011; Rabita et al., 2011). This indicated that changing the COM vertical displacement affected the vertical stiffness more than changing the peak vertical force (Cavagna et al., 1988; Morin et al., 2007; Matt; Brughelli and Cronin, 2008). Meanwhile, a positive correlation was found between the vertical displacement of COM and RE (Halvorsen et al., 2012; Tartaruga et al., 2012;

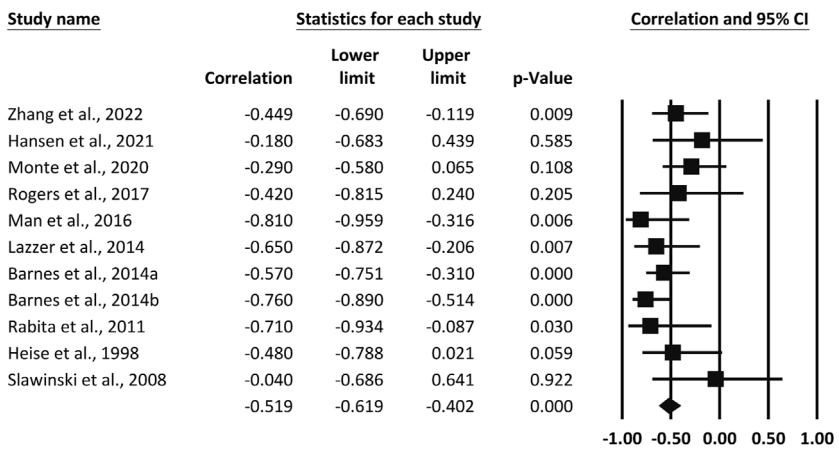

FIGURE 2
Forest plot of the correlation between vertical stiffness and running economy. The superscripted letter in column 1 refer to speed or gender items assessed in the same study.

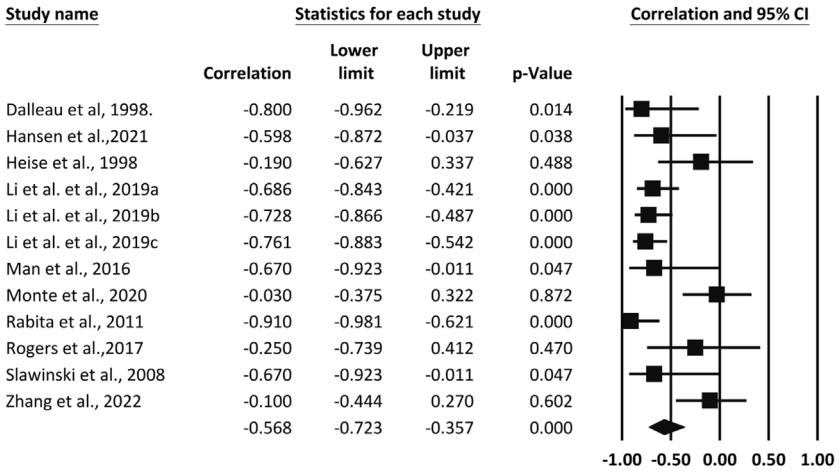

FIGURE 3
Forest plot of the correlation between leg stiffness and running economy. The superscripted letter in column 1 refer to speed or gender items assessed in the same study.

Folland et al., 2017). Given the absorption and release of elastic energy, greater vertical oscillation necessarily involves more work against gravity (Moore, 2016; Folland et al., 2017). The first study comparing the biomechanical characteristics of RE in runners at different levels showed that elite runners had fewer vertical oscillations, which were more symmetrical (Cavanagh et al., 1977). Farley et al. (1998) suggested that increasing vertical stiffness by reducing the COM displacement allows the spring-mass system to recoil in a shorter period, facilitating faster absorption and generation of kinetic energy during ground contact. In addition, greater vertical stiffness helped to resist flexion of the lower extremity joint during the support phase. It increased the rate of force generation during the centrifugal (centripetal) phase, which enhanced the storage and utilization of elastic energy during SSC (Matt Brughelli and Cronin, 2008).

The study found that maintaining a high whole-body vertical stiffness minimizes oxygen consumption, and smaller heterogeneity enhances the certainty of the results. Meta-regression results show that the correlation between vertical stiffness and RE is less susceptible to the effects of runner level and speed. Several studies have demonstrated that

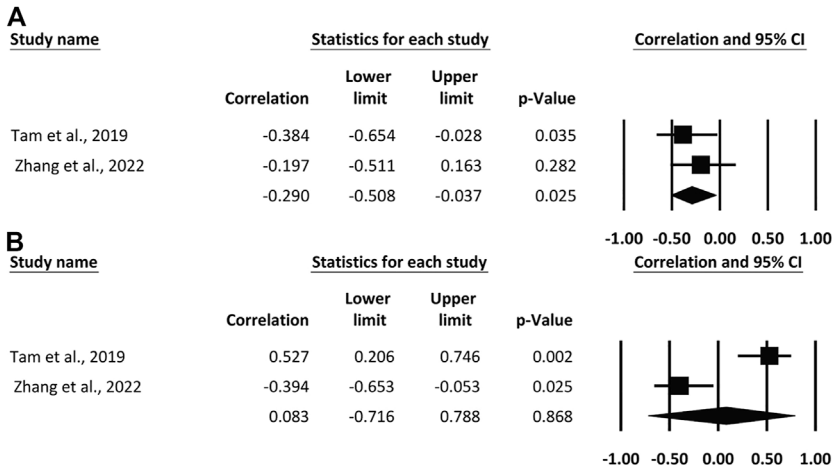

FIGURE 4
Forest plot of the correlation between knee **(A)** and ankle stiffness **(B)** and running economy.

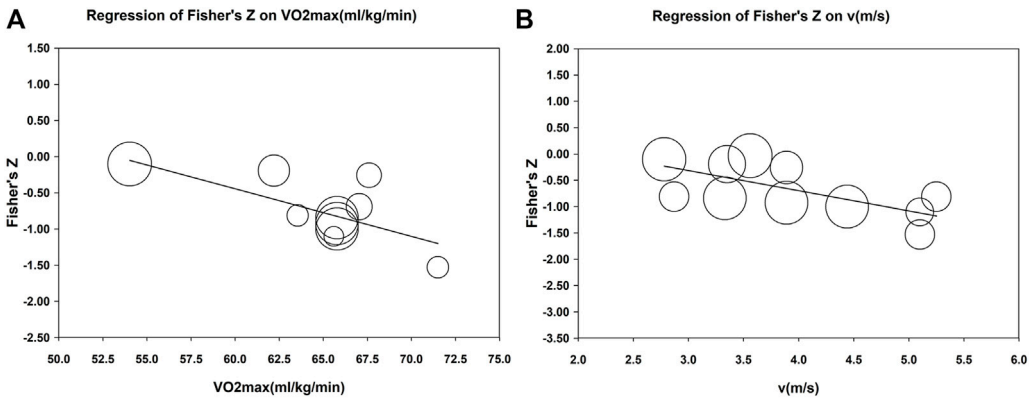

FIGURE 5
Meta-regression of VO_{2max} **(A)** and test velocity **(B)** with correlation coefficients of leg stiffness and RE. The size of each circle is proportional to the study's weight.

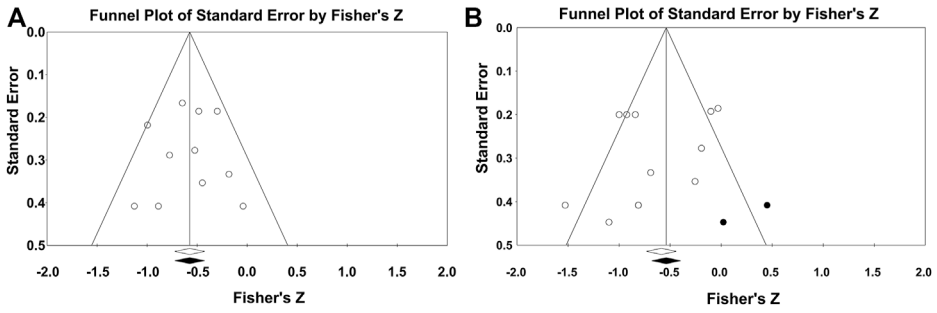

FIGURE 6
Vertical stiffness **(A)** and leg stiffness **(B)**, funnel plot with running economy correlation.

trained runners who run to exhaustion maintained initial vertical stiffness (Hunter and Smith, 2007; Morin et al., 2011) and even increased it significantly in 24-h ultra-long runs (Morin et al., 2011).

Leg stiffness and running economy

Our results found the same negative correlation between leg stiffness and RE during the submaximal running, suggesting that leg stiffness is an essential factor influencing RE. Leg stiffness is an essential parameter in regulating running mechanics, which can maintain a stable running gait in humans and animals (Seyfarth et al., 2002). The following equation can calculate the leg stiffness:

$$k_{leg} = \frac{F_{max}}{\Delta L},\ \Delta L = \Delta y + L(1 - \cos\theta),\ \theta = \sin\left(v \cdot \frac{tc}{2L}\right)$$

where v is running velocity and tc is contact time. Leg length is symbolized as L, which comes into contact with the ground at an angle of θ (McMahon and Cheng, 1990).

Several studies support the result of a significant negative correlation in the random effects of this study. Scholars began to focus on the possibility that RE may be related to the time course of supporting body weight, for example, ground contact time (tc). Kram et al. (1990) reported that the cost of supporting the animal's weight and the time course of generating this force determines the cost of running. Because the longer tc allows the lower-extremities to generate propulsive force over a longer period of time when in contact with the ground, reducing energy costs (Kram and Taylor, 1990). Notably, many studies on human running showed a significant positive correlation between tc and RE (Nummela et al., 2007; Barnes et al., 2014; Di Michele and Merni, 2014; Mooses et al., 2021). This was likely due to less time required for braking to decelerate the body's forward motion (Nummela et al., 2007; Kong and de Heer, 2008; Mooses et al., 2021). Meanwhile, a negative correlation between leg stiffness and tc was demonstrated in other studies (Morin et al., 2007; Hayes and Caplan, 2014; Santos-Concejero et al., 2014; Man et al., 2016). Morin et al. (2007) found that the change in tc can account for 90% of the change in leg stiffness. This means that reducing tc increases leg stiffness (Man et al., 2016). This may be due to the high preactivation of the calf muscles which then increases the sensitivity of muscle spindle potentiating stretch reflexes to enhance musculo-tendon stiffness and to improve the RE (Santos-Concejero et al., 2014; Mooses et al., 2021). Moreover, Moore et al. (2019) showed that the relationship between leg stiffness and RE is not linear and that there is an identifiable optimum for it, as 90% of runners can keep their optimal metabolic cost within 5% at a self-selected leg stiffness (Moore et al., 2019). Unfortunately, the optimal range of leg stiffness values for endurance runners is still unknown.

The results of the meta-regression explained the high heterogeneity of the correlation between leg stiffness and RE. The results showed that the correlation between leg stiffness and RE increased with the speed and runner VO_{2max}. Previous studies showed that novice runners have higher oxygen consumption than trained runners, while tc is longer (Williams and Cavanagh, 1987; de Ruiter et al., 2014). Bitchell et al. (2019) state that although leg stiffness was unaffected by physiological training status, untrained runners had difficulty maintaining consistent leg stiffness during running. This variability is likely related to increased oxygen costs. Experienced and well-trained runners optimize elastic energy storage and release more rationally to minimize metabolic costs (Lussiana et al., 2019; Moore et al., 2019).

There has been no consensus whether speed affects leg stiffness in previous studies (Matt Brughelli and Cronin, 2008). Our meta-regression results confirmed that speed enhanced the correlation between leg stiffness and RE. The greater the speed, the more pronounced the favorable effect of leg stiffness on RE. Lai et al. (2014) reported that the energy stored in the elastic deformation of the tendon is better utilized at higher speeds and that the positive work done by the elastic strain energy on the tendon-muscle unit is lower at slower running speeds than at faster running speeds. The results of this study validate previous conjectures that there is indeed a "U" shaped relationship between leg stiffness and RE and that training level and running speed will determine whether a runner's energy cost is near the bottom of the curve (Zhang et al., 2022).

Joint stiffness and running economy

The torsional spring model can be utilized to estimate the joint stiffness values of the main joints of the lower extremity during vertical and horizontal movements (Latash and Zatsiorsky, 1993). The following equation can calculate joint stiffness:

$$k_{joint} = \frac{M}{\Delta\alpha}$$

where M denotes the deformation torque and $\Delta\alpha$ is the deformation angle.

Measurement of joint stiffness, particularly of the ankle and knee joints, provides greater insight into the respective contributions of the joints to the overall stiffness of the lower extremity (Maloney and Fletcher, 2021). The combination of knee and ankle stiffness provides the best correlation for leg stiffness when exploring the variation in leg stiffness (Lorimer et al., 2018). Unfortunately, there is only two evidences supporting the relationship between joint stiffness and RE. Tam et al. (2019) found that high knee and less ankle stiffness were associated with better RE. The results of the meta-analysis

showed a negative correlation between knee stiffness and RE, though it was not significant.

Joint stiffness mainly depends on the level of activation of the muscles around the joint (Farley et al., 1998). The most economical runners rely on greater muscle activation (Tam et al., 2019). Kyröläinen et al. (2001) found that muscle co-activation around the knee and ankle joints during running increases joint stiffness, which was associated with better RE. Tam et al. (2017) also showed that during the pre-activation and ground contact, rectus femoris (RD):biceps femoris (BF) co activation ratio was positively correlated with the knee, but not the ankle joint. Pre-emptive neuromuscular joint control decreases the need for corrective muscle activation at and after ground contact, where loading forces are applied to joints stiffness (Tam et al., 2019). This suggested that greater knee stiffness allowed more energy to be stored in the leg spring through co-activation of the agonist and antagonist muscles and provided coordination of the ankle and hip joints (Jin and Hahn, 2018).

Unlike walking, running requires more bracing phase energy generation (Jin and Hahn, 2018). Meanwhile, the ankle joint does more positive work in the support phase than the knee (Jin and Hahn, 2018). During landing, the knee and hip joints absorb more of the impact, which requires higher ankle stiffness and the joint muscle-tendon pull reflex, resulting in more energy demand (Jin and Hahn, 2018). Previous studies have shown that increased gastrocnemius-anterior tibialis activation is associated with high ankle stabilization stability and energy sparing (Kyröläinen et al., 2001; Fletcher and MacIntosh, 2017; Tam et al., 2017; Tam et al., 2019).

The reason for the opposing results in the two studies may stem from differences in landing patterns and running speeds (Tam et al., 2018; Zhang et al., 2022). The forefoot landing ankle stiffness used by the elite runners was lower and performed more negative work (Hamill and Gruber, 2017). This suggests that a high supple joint stiffness absorbs the energy of landing impact to a greater extent. The hindfoot landing has high ankle stiffness and performs less negative function (Hamill and Gruber, 2017). However, as speed decreases and running distance increases, some runners tend to land on their hind feet (Hasegawa et al., 2007). Additionally, Weir et al. (2020) found that habitual hindfoot runners showed a decreased ankle stiffness as the running time increased. In conclusion, the current limited evidence makes the results uncertain and future studies are necessary to address the comparison of ankle stiffness contributions at different running patterns and speeds.

Since the phase shift for the moment-displacement curve of the hip commonly exceeds 10% (Farley and Morgenroth, 1999; Kuitunen et al., 2011; Maloney and Fletcher, 2021), it is often excluded from the comparison. As a result, many studies about joint stiffness have only focused on the knee and ankle joints, and little attention has been paid to hip stiffness. Jin and Hahn (2018) observed that the hip joint plays an important role in energy absorption during the swing phase of running. However, the relationship between hip stiffness and endurance in sports performance is unknown.

Limitations and prospects

Several limitations were identified in this study. First, the number of studies included in this study was too small, and the Pearson correlation coefficient still has some limitations in expressing the relationship between lower extremity stiffness and RE. Second, the vast majority of subjects in the study were male, with only 11% of females participating. In one of the articles used in this study, women showed a greater correlation than men. Although RE may not differ by gender (Besson et al., 2022), there are differences in the biomechanics of running (Barnes et al., 2014; Besson et al., 2022). For example, women have faster stride length, shorter *tc*, and faster time to peak (Nelson et al., 1977; Williams et al., 1987; Besson et al., 2022). As a result, women have higher leg stiffness than men (Barnes et al., 2014). Besson et al. (2022) stated that male and female runners optimize their running patterns to suit their gender characteristics. In addition, women may have physiological and technical advantages in ultra-endurance events (Tiller et al., 2021). As a result, future researchers need to pay more attention to the relationship between lower extremity biomechanics and energy in runners of different genders.

The above conclusion are based on SMM. This simple model can reproduce the motion and forces of running remarkably well. Schroeder and Kuo (2021) proposed a running model based on SMM that combines elasticity with active actuation and passive dissipation (Actuated Spring-mass model). This model included two types of dissipation, one dissipating for collision and hysteresis losses, and the other including hysteresis to simulate imperfect energy return of tendons and other series elastic tissues (Schroeder and Kuo, 2021). However, the model is not perfect either, for example, it neglects a swing leg, whose active motion may also cost energy (Doke et al., 2005). In the future, more sophisticated models should be adopted to simulate the energy expenditure during running.

Finally, we discussed the relationship between each of the three stiffnesses and RE. A negative correlation was obtained for all except ankle stiffness. However, the consistency results may lead to the mistaken belief that they are independent and interchangeable. They act together on the body in motion and support the body mass. There is a close link between vertical, leg and joint stiffness, for example, the stiffness of the leg spring is influenced by the stiffness of the three joints (hip, knee, and ankle) (Struzik et al., 2021). Therefore, in the future, it is necessary to continue to explore the potential relationships between vertical, leg, and joint.

Conclusion

This meta-analysis found a significant negative association of vertical, leg, and knee stiffness with RE. Furthermore, the extent

of this correlation was related to speed and the runner's VO_{2max}. This suggests that experienced runners make more rational use of elastic energy to minimize energy costs at faster speeds. Some scholars have suggested that there is a weak negative correlation between knee stiffness and oxygen consumption, but it is not clear how much ankle stiffness runners need. Many factors affect stiffness during running, and these factors are influenced by different tasks, populations, and gender. The optimal stiffness required for a runner needs to be further investigated. The results of this study may, to some extent, inform future training and research.

Data availability statement

The original contributions presented in the study are included in the article/Supplementary Material, further inquiries can be directed to the corresponding author.

Author contributions

BL and ZR designed the research. BL and WX conducted the searches and screening process. BL and WX completed the full-text screening. BL, WX, JY, and FH assessed methodological quality. BL, FY, and FH extracted the data, which were checked by JW. BL performed the statistical analysis and interpreted it. BL wrote the manuscript with critical input from ZR. ZR and QS read and approved the final manuscript.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Effects of kinesio taping therapy on gait and surface electromyography in stroke patients with hemiplegia

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Background: The application of Kinesio Taping (KT) on the lower extremity of stroke patients can improve the quality of somatosensory information by activating lower extremity muscles involved in postural control. Gait analysis and surface electromyography (SEMG) are valuable in assessing the motor ability of the lower extremities.

Objective: This study aimed to investigate the effects of KT therapy on gait and SEMG in stroke patients with hemiplegia.

Methods: Twenty-one stroke patients were included in the study. KT was applied to the lower extremities of the hemiplegic side. Quantitative gait parameters were measured by a gait analysis system (IDEEA, by MiniSun, United States) and activation of the lower extremity muscles were evaluated by the SEMG (Trigno™ Wireless Systems, Delsys Inc., United States) before and after taping. Step length, stride length, pulling acceleration, swing power, ground impact, and energy expenditure were used to evaluate when patients walk as usual. SEMG signals were collected from the anterior bilateral tibialis (TA) and the lateral gastrocnemius (LG). The root mean square (RMS) value was used to assess muscle activity. SEMG signals were examined before and after KT treatment in three different locomotor conditions of the patients: walking at a natural speed, walking with a weight of 5 kg, dual-tasking walking (walking + calculation task) while carrying a weight of 5 kg. The calculation task was to ask the patients to calculate the result of subtracting 7 from 100 and continuing to subtract 7 from the resulting numbers. Comparisons between two normally distributed samples (before and after KT treatment) were evaluated using the two-tailed, paired Student's *t*-test.

Results: Stride length (0.89 ± 0.19 vs. 0.96 ± 0.23; $p = 0.029$), pulling acceleration (0.40 ± 0.21 vs. 1.11 ± 0.74; $p = 0.005$), and swing power (0.42 ± 0.24 vs. 1.14 ± 0.72; $p = 0.004$) improved in the hemiplegia side after KT treatment. The RMS value of TA SEMG signals in the limbs on the hemiplegia side decreased after KT treatment during dual-tasking walking carrying a weight of 5 kg (3.65 ± 1.31 vs. 2.93 ± 0.95; $p = 0.030$).

Conclusion: KT treatment is effective in altering gait and SEMG characteristics in stroke patients with hemiplegia.

KEYWORDS

stroke, kinesio taping, sEMG, gait analysis, rehabilitation

Introduction

Stroke is a common cerebrovascular event that can result in disabilities including motor, sensory, visual, and cognitive impairments (Ekker et al., 2018). About 70% of stroke survivors exhibit a degree of motor dysfunction that affects daily activities of the patient, social participation, and quality of life (Schröder et al., 2019). Furthermore, in recent decades, an increase of up to 40% has been reported in the incidence of stroke in young adults (Ekker et al., 2018). These young patients have a long-life expectancy after stroke, and the cost of long-term care is a significant burden on healthcare systems. Recovery of motor function on stroke patients is a long and complicated process that requires patients to undergo extensive rehabilitation therapy that involves frequent and regular exercises that match their impairments (Yang et al., 2015). Walking dysfunction is the most common problem in post-stroke patients, and involves an inability to use the ankle dorsiflexor, abnormal gait, and an increased risk of falls due to foot drop (Yang et al., 2015). Falls in stroke patients are closely related to abnormal activation of lower limb muscles (Beyaert et al., 2015; Peishun et al., 2021).

Current common treatment methods for walking rehabilitation include acupuncture, exercise therapy, physiotherapies such as functional electrical stimulation (FES), ankle-foot orthotics (AFO), and kinesio taping (KT) methods (Sheng et al., 2019a). KT is an elastic adhesive tape attached to the surface of the body and is currently used to promote lymphatic circulation, ease pain, provide mechanical support, and improve proprioception (Ekiz et al., 2015). FES can activate the muscles that dorsiflex the ankle and offer an alternative to this treatment for facilitating motor restoration (Sabut et al., 2010). However, electrode placement and instrument manipulation of many FES devices are complex. AFO exert limited effects on walking ability due to increased fatigue and restricted ankle movement (Prenton et al., 2018). In contrast, KT has long been used to strengthen weakened muscles, control muscle tone, improve the active range of motion, balance, functional use, and gait ability as a cost-effective treatment (Park and Bae, 2021). Despite the different treatments available, the mechanisms underlying lower limb muscle activation in stroke patients associated with falls warrants further study.

KT is widely used in the treatment of stroke patients, but the regulation of muscle activation after KT treatment is still unclear. At present, there have been few studies that have objectively evaluated the therapeutic effects of KT on gait stability in stroke patients (Choi et al., 2013; Babyar et al., 2014; Yang et al., 2015). Recently, gait analysis was introduced the ability to measure gait parameters accurately and precisely, allowing clinicians to obtain patient gait information quickly and easily (Gardner et al., 2007). Gait analysis is a well-established tool for the quantitative assessment of gait disturbances, which is helpful to assess the effect of rehabilitation and monitor the disease progress (Baker et al., 2016).

Surface electromyography (SEMG) has the advantages of non-invasion, real-time, and multi-target measurement, and is a method that has received growing attention due to its ability to quantitatively analyze neuromuscular activity in static and dynamic motion states (Rasool et al., 2017). SEMG is a biological electrical signal of the neuromuscular system that is guided and recorded by electrodes placed on the muscle surface (Frigo and Crenna, 2009). SEMG can be used for static muscles, and can observe changes in muscle activity during numerous sports activities (Qie et al., 2020). For example, SEMG has been used to assess normal and abnormal muscle activation in stroke patients and in some patients with sports-injuries, such as those with anterior cruciate ligament injury, to guide rehabilitation strategies (Papagiannis et al., 2019).

KT is an effective treatment for pain caused by sports injury or other diseases, such as osteoarthritis, chronic low back pain, chronic skeletal muscle pain, and delayed onset muscle soreness (Kalinowski and Krawulska, 2017; Hazar Kanik et al., 2019; Rahlf et al., 2019). KT can also improve active range of motion (ROM) and limb function in patients with sports injury (Merino-Marban et al., 2013). Activity during extension movement on taping is conducive to promoting local circulation and improving ROM. In the treatment of stroke patients, the application of KT has been evaluated on shoulder, trunk, upper limb, lower limb and ankle movements, but there have been no studies on the application of KT therapy on the calf and ankle movements simultaneously. SEMG is commonly used to assess muscle function in stroke patients, but is rarely used to assess muscle activation in multiple motor states. In this study, SEMG was used to assess electrophysiological changes in calf muscle activation in stroke patients under three different motor conditions.

Sheng et al. (2019b) found signifcant improvement in stride length, stance phase and swing phase of stroke patients after KT therapy. They suggested that KT may help to improve posture control and instantly produce immediate efects on walking. Jin et al. showed that the study group had significant increases in RMS of vastus medialis SEMG signals after 16 weeks of

TABLE 1 Characteristics of the subjects.

Characteristics	Median
Gender (Male/female)	15/6
Age (yrs)	51 (42–57)
Height (cm)	170 (164–172)
Weight (kg)	71.5 (60–78.5)
BMI (kg/m^2)	24.91 (23.42–27.73)
Type of stroke (Hemorrhage/infarction)	5/16
Affected side (Left/right)	10/11

Baduanjin training (Jin et al., 2017). They also pointed out that RMS of SEMG signals can be used to analyze the recruitment of muscle fibers during contraction. We hypothesized that KT treatment would improve the stride length in stroke patients. The RMS of hemiplegic side TA SEMG signals would increase and RMS of affected side LG SEMG signals wouldn't change significantly while walking in different locomotor conditions.

Materials and methods

Subjects

Twenty-one post-stroke patients with hemiplegia recruited from April 2021 to December 2021 were enrolled in this study. Patients who met the following criteria were included: 1) aged > 30 years; 2) diagnosed with stroke; 3) had a 3–6 months course of stroke; 4) the Brunnstrom stages are at least four; 5) able to walk without assistance; and 6) had foot drop after stroke on the hemiplegic side. Exclusion criteria were as follows: 1) patients with cognitive impairment unable to complete experimental procedures; 2) patients with a history of other neurological diseases or disorders, lower extremity surgery or fracture; or 3) patients exhibiting allergy to KT. At baseline, descriptive variables for each patient, including height, weight and Body Mass Index (BMI), were recorded. All patients were in the middle and late stages of stroke recovery, and their activities of daily living were basically self-care (Barthel index > 60 points). The general characteristics of the subjects are shown in Table 1. The nature and purposes of this study were explained to all participants and all signed an informed consent form prior to participation in the study. The Number of the ethical approval letter is 2021-G 001-01.

Taping intervention

Patients underwent routine rehabilitation, including complete training in the hemiplegic limb, exercise training in walking function. All KT treatments were performed by the same qualified physical therapist on the hemiplegic limb of the patient. The area to be taped was cleaned with an alcohol swab prior to KT application. The patients were placed in a supine position during the taping, with their hip, knee, and ankle joints in a neutral position (Rojhani-Shirazi et al., 2015). A kinesio I-shaped strip was placed from the upper end of the external side of the tibia downward along the anterior tibialis muscle and ending at the ankle joint. Two other kinesio I-shaped strips were fixed at the ankle joint in the shape of "x", where the first was stretched 20-cm from the heel along the Achilles tendon while the other extended from the Achilles tendon to the medial and external malleolus (Figure 1). The pulling force of each kinesio strip was 10% of the maximum tensile length of the tape (Morris et al., 2012; Choi et al., 2013).

Testing procedures

All evaluations were performed by a qualified physical therapist. Before and after taping, the gait function of the patients was measured using an Intelligent Device for Energy Expenditure and Activity (IDEEA, by MiniSun, United States) device. It is capable of identifying 35 activities and postures and can provide an estimate of energy expenditure when incorporated with basic subject anthropometry (Whybrow et al., 2013) and is capable of both detecting and describing motion as specific activities, including sitting, walking, or stair use, and defining temporal-spatial gait parameters (Mackey et al., 2008). While patients performed the walking exercise, the IDEEA device was used to measure gait parameters including single-limb support time, double-limb support time, swing phase duration, cycle duration, cadence (number of steps per minute), stride length (the distance between the heel points of two consecutive footprints of the same foot), pulling acceleration, swing power, ground impact, foot fall (neuromuscular and skeletal control of the limbs during the end of swing phase), and energy expenditure (EE) (Figure 2). SEMG was measured using a DELSYS wireless dynamic EMG tester (Trigno™ Wireless Systems, Delsys Inc., United States). The bilateral TA and LG SEMG signals were collected and analyzed as these are widely used to evaluate the function of the lower extremity. Subjects walked back and forth under three different locomotor conditions on a 10-m flat floor and the SEMG signals were collected and transmitted *via* a wireless Bluetooth connection. Each parameter was averaged over 20 gait cycles to obtain a representative EMG profile for each muscle. SEMG signals were collected from the TA and LG (Figure 2) under three different locomotor conditions: walking at natural speed, walking holding a weight of 5 kg, and dual-tasking walking (walking + calculation task) holding a weight of 5 kg. For the calculation task, patients were asked to calculate the result of subtracting 7 from 100 and to subtract 7 from the resulting number, and so on. Evaluations were conducted before and after KT taping immediately. All

FIGURE 1
KT taping methods.

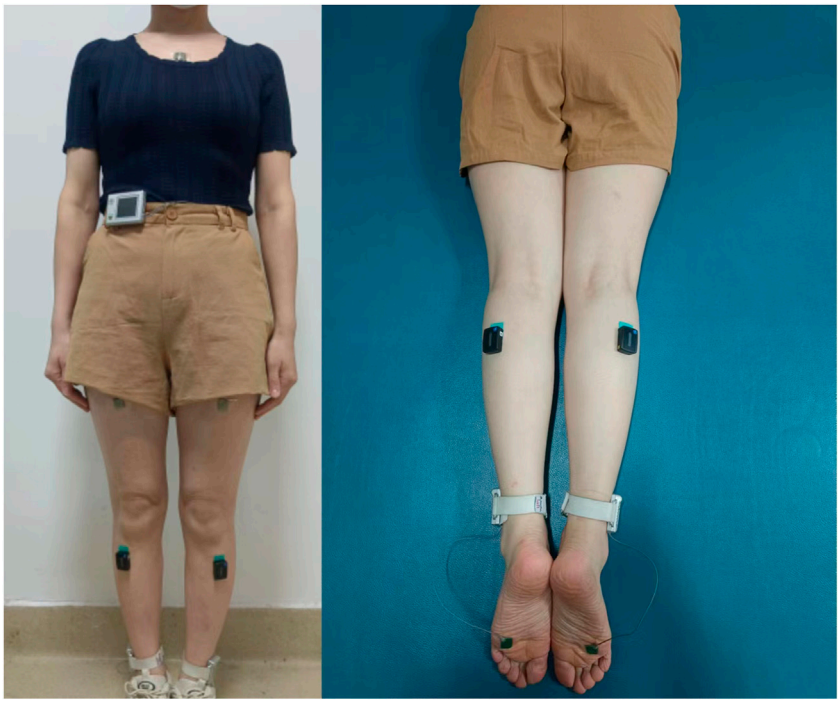

FIGURE 2
The IDEEA system worn by a patient and the locations of the SEMG electrode sites on the TA and LG.

TABLE 2 Comparison of the gait parameters before taping and after taping for patients (N = 21) walking at a normal pace.

Variables	Before taping (M ± SD)	After taping (M ± SD)	p-value
Step length (m)	0.44 ± 0.10	0.48 ± 0.12	0.190
Stride length (m)	0.89 ± 0.19	0.96 ± 0.23	0.029*
Cadence (steps/min)	87.38 ± 23.04	92.06 ± 20.31	0.138
Speed (steps/sec)	0.69 ± 0.30	0.75 ± 0.30	0.122
Single support phase (%)	38.62 ± 4.04	36.31 ± 4.63	0.255
Double support phase (%)	15.68 ± 6.66	16.55 ± 7.33	0.715
Pulling acceleration (G)	0.40 ± 0.21	1.11 ± 0.74	0.005*
Swing power (G)	0.42 ± 0.24	1.14 ± 0.72	0.004*
Ground impact (G)	0.80 ± 0.45	0.85 ± 0.42	0.242
EE (kcal/m)	3.24 ± 1.10	3.19 ± 0.95	0.843

*Indicates significant differences ($p < 0.05$).

measurements were collected twice for each patient (Sheng et al., 2019a).

Signal processing

The SEMG signals were processed and analyzed using EMGworks Analysis and MATLAB R2020a (MathWorks, 3 Apple Hill Dr, Natick, MA 01760-2098). The sampling frequency was 1,200 Hz. The root mean square (RMS) value was analyzed and used to evaluate muscle activity. TA and LG muscles were intermittently activated during the test. SEMG data was collected continuously throughout the test. The signals for each dorsal extension and plantar flexion of the ankle were isolated for further analysis. We combined SEMG data from 20 gait cycles using SEMG analysis software. SEMG synchronization points were subjected to analysis using MATLAB software after noise elimination. First, the raw SEMG signals were digitally zero-phase filtered by a second-order Butterworth bandpass filter (20–450 Hz) and then they were full-wave rectified. Second, the RMS values of the full-wave rectified signals were computed within a 50-ms time window. The baseline SEMG activity was determined to be six times the standard deviation of the first 50 ms of the RMS values, where there was no contraction in the dorsal extension and plantar flexion of the ankle muscles. RMS values greater than the baseline were detected and used to determine the beginning and ending of the dorsal extension and plantar flexion of ankle epochs. The difference between the indices of these values was calculated. The point where the difference between the two indices is greater than 250 ms was identified as the end of an epoch. In accordance with de Jesus et al. (2010), the rectified SEMG signals in the identified epochs greater than the baseline values were defined as effective myoelectric activation (EMA) of the muscles.

Statistical analysis

All the data was analyzed in SPSS software (Version 26.0, IL, United States). Data are reported as mean ± standard deviation (M ± SD). The Shapiro–Wilk normality test was performed to examine the normality distribution of the data. Comparisons between two normally distributed samples (before and after KT treatment) were evaluated using the two-tailed, paired Student's t-test. A p-value < 0.05 was considered statistically significant.

Results

Gait measurement results

Stride length [before: 0.89 ± 0.19 (m); after: 0.96 ± 0.23 (m); p = 0.029], pulling acceleration [before 0.40 ± 0.21 (G); after: 1.11 ± 0.74 (G); p = 0.005], and swing power [before: 0.42 ± 0.24 (G); after: 1.14 ± 0.72 (G); p = 0.004] all increased after taping when patients were performing walking activities (Table 2).

Surface electromyography measurements

SEMG was measured during three walking modes (walking, walking carrying a weight of 5 kg, and dual-tasking walking carrying a weight of 5 kg and calculation tasks simultaneously) for all patients. The RMS of the hemiplegic side TA SEMG signals significantly decreased (before: 3.65 ± 1.31; after: 2.93 ± 0.95; p = 0.030) after KT treatment during dual-tasking walking carrying a weight of 5 kg but not for other modes of walking (Table 3).

Discussion

In this study, we found that KT could increase stride length, pulling acceleration, and swing power ($p < 0.05$) in affected limbs, improve calf muscle strength, and improve gait stability when patients walked at natural speed. This result is consistent with previous hypothesis that KT treatment would improve the stride length in stroke patients. Furthermore, we also found the RMS value of the affected side TA SEMG signals were different from LG muscles. The RMS value of affected side LG SEMG signals increased after KT taping, but there was no statistical significance ($p > 0.05$). The RMS value of the affected side TA SEMG signals decreased ($p < 0.05$) after KT treatment when the patients performed dual-tasking walking carrying a weight of 5 kg. This result is inconsistent with previous hypothesis that the

TABLE 3 Comparison of affected-side muscles SEMG signals before taping and after taping when patients performed three distinct walking modes.

Variables	TA (M ± SD)			LG (M ± SD)		
RMS (V)	Before taping	After taping	*p*-value	Before taping	After taping	*p*-value
Walking at a natural speed	3.71 ± 1.37	3.16 ± 1.13	0.086	2.59 ± 0.86	2.85 ± 1.25	0.333
Walking with a weight of 5 kg	3.64 ± 0.93	3.70 ± 1.50	0.782	2.62 ± 1.02	2.91 ± 1.29	0.285
Dual-tasking walking with a weight of 5 kg	3.65 ± 1.31	2.93 ± 0.95	0.030*	2.47 ± 0.95	2.87 ± 1.28	0.233

*Indicates statistical significance ($p < 0.05$).

RMS of hemiplegic side TA SEMG signals would increase and RMS of affected side LG SEMG signals would decrease while walking in different locomotor conditions.

Gait parameters such as stride length, pulling acceleration and swing power were improved after KT therapy. We attributed these findings to the fact that strengthening the TA muscle and mechanical support of the KT contributes to better ankle control. Kase et al. have reported that KT increases muscle activation through the following two mechanisms (Ekiz et al., 2015). Firstly, KT stimulates cutaneous receptors by tactile stimulation and increases sensory input to recruit more motor units during the most vigorous contraction of the muscle (Chang et al., 2010). Secondly, KT increases the subcutaneous volume and blood flow (Akbaş et al., 2011). In addition, studies have shown that walking ability and efficiency of the patient improved immediately after the KT treatment (Ekiz et al., 2015; Choi et al., 2016; Sheng et al., 2019b). Lee and Bae (2021) demonstrated the short-term effect of application of lower-leg KT according to the proprioceptive neuromuscular facilitation (PNF) pattern increased the gait ability of chronic stroke patients with foot drop. KT application has also been found to have a more positive effect on stance phase duration than McConnell taping in patients with stroke (Sung et al., 2017). Park et al. compared the immediate effects of talus stabilization taping (TST) with those of KT on the ankle dorsiflexion passive range of motion (DF-PROM), static balance ability, the Timed Up and Go (TUG) test, and fall risk in patients with chronic stroke (Park and Lee, 2019). Further, the TUG results decreased significantly in the KT group. These results are consistent with a previous study (Sheng et al., 2019a). Sheng et al. (2019a) demonstrated that the ankle KT intervention significantly improved the 10-m walking test, the TUG results, the stride length, the stance phase, and the swing phase in patients with a foot drop after stroke. The application of viscoelasticity and continuous mechanics of KT can also support and stabilize muscles and joints, deepen sensory input and promote circulation that reduces pain and swelling (Koseoglu et al., 2017).

In the study, the RMS value of the affected side TA SEMG signals decreased ($p < 0.05$) after KT treatment in stroke patients while performing multiple tasks synchronously. We could speculate that when patients performed multitask walking synchronously, the activation of the affected side TA muscle decreased. We considered that the patient should pay closer attention on multitasking, which affected the precision of the walking movement (de Barros et al., 2021). Walking ability is related to cognitive functions such as executive function and attention (O'Brien and Holtzer, 2021; Ueda et al., 2018). Executive function is a series of high-level functions involved in the processing of information from the posterior cortical sensory system to the anterior cortex to produce the corresponding movements, including the intention or start of action (walking), decision-making and control behavior. Attention is a dynamic function driven by sensory perception that requires selecting a primary stimulus for a particular behavior (walking) while ignoring unnecessary and irrelevant stimuli. Studies have also shown that due to neural damage affecting postural control, greater cognitive resources must be mobilized during walking, leading to a high cognitive load in stroke patients when dealing with dual tasks. Postural control motor dysfunction also increases the demand for limited attention resources (Yang et al., 2018; Koren et al., 2022).

The cognitive function of stroke patients tends to decline. Cognitive-motor and motor dual tasks play important roles in daily life, such as walking while talking, using a mobile phone, carrying a bag, or watching traffic (Yang et al., 2007). Previous studies have indicated that performing two tasks simultaneously can negatively impact on gait performance. Dual task interference that affects gait performance has been observed not only in healthy subjects, but also in subjects with neurological disorders (Yang et al., 2007). In stroke-injured individuals, reductions in speed, cadence, and stride length, as well as increases in stride time during cognitive-motor dual tasking have been reported. In addition, stroke subjects have more difficulty performing dual motor tasks compared to healthy adults (O Brien and Holtzer, 2021). Diminished capacity for dual task performance and reduced ability to adapt to changing environments may limit the ability of individuals with stroke to return to the community. Consequently, improving walking ability in dual task situations is an important goal, especially for subjects experiencing chronic stroke with limited ambulation in the community setting (Lu et al., 2015).

Furthermore, gait analysis and SEMG technology were used to provide an objective and quantitative method for evaluating the effect of KT therapy. Approximately 70% of patients with hemiplegia recover their ability to walk after these therapies but

retain abnormal gait and problems such as foot drop and pronation (Gandolla et al., 2018). These abnormal movement patterns affect stability, walking safety, and consume more energy than a regular gait (Cho et al., 2020). It remains a difficult problem to improve gait ability in the legs of patients with hemiplegia in rehabilitation therapy (Park and Lee, 2016). Furthermore, an objective and quantitative evaluation of the rehabilitation effect is also the trend and requirement of precision rehabilitation therapy (Babyar et al., 2014; Wonsetler and Bowden, 2017; Lerma Castaño et al., 2020). However, most studies use scales and experimental tests, such as the modified Ashworth Scale (MAS), the Stroke-Specific Quality of Life Scale (SS-QLS), and the 10-m walk test (10 MWT); these assessments can be affected by subjective factors (Mohan et al., 2021). The combination of gait analysis and SEMG can quantitatively evaluate the amount of rehabilitation experienced by patients and thus provide an objective reference for rehabilitation programs (Zhu et al., 2019; Peishun et al., 2021).

Moreover, we observed improvement in the stability of the ankle joint and improvement of the high tension of the extensor muscle when walking using a taping formula based on the plasticity theory of the central nervous system (Park and Bae, 2021). As such, continuous sensory input was similar to the continuous correction of deviation by a therapist (Hu et al., 2019). Compared to traditional one-to-one rehabilitation treatment, patients detect correct motion guidance more directly when using KT (Santos et al., 2019). Clinicians can also observe a patient's gait and muscle activation with a gait analyzer and SEMG, which allows for adjustments to the rehabilitation treatment plan in real time (Choi et al., 2016).

This study demonstrates that KT treatment can improve gait in stroke patients with hemiplegia when patients do not perform multiple tasks simultaneously. However, some limitations to the present study should also be considered. First, this study did not conduct a hierarchical comparison study that included multiple types of stroke or tested different KT taping methods as the etiopathological mechanism(s) and the clinical presentation of stroke is complex (Lee et al., 2017). We also need to further explore changes in brain function during multitasking. Furthermore, the lasting effects of the application of KT therapy are unknown due to the small number of samples and the lack of long-term follow-up observation; thus, additional research is necessary to establish the duration and efficacy that KT provides.

Conclusion

KT therapy is effective in improving gait ability and muscle function among stroke patients with hemiplegia, KT therapy helps prevent falls and promotes recovery among stroke patients with hemiplegia.

Data availability statement

The original contributions presented in the study are included in the article/supplementary material, further inquiries can be directed to the corresponding authors.

Ethics statement

The studies involving human participants were reviewed and approved by the Scientific Research Ethics Committee of the Binzhou Medical University Hospital. The patients/participants provided their written informed consent to participate in this study.

Author contributions

ZC contributed to acquisition of data, data analysis and drafting of manuscript. ML contributed to acquisition of data and design of study. HC contributed to acquisition of data. XW contributed to giving suggestions for revision about the manuscript. FC contributed to the conception of the manuscript. WL contributed to putting forward the research concept, modify the manuscript and approve the final version of manuscript. ZC and ML contribute equally to this article. WL and FC are co-corresponding authors.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Analysis of plantar impact characteristics of walking in patients with flatfoot according to basic mechanical features and continuous wavelet transform

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Flatfoot is a common foot deformity, and the collapse of the arch structure affects the foot cushioning during walking. A growing number of scholars have found that the analysis of the impact force should be combined with both the magnitude and the frequency of the impact force. Therefore, the aim of this study is to investigate the plantar impact characteristics of flatfoot patients and normal foot people at different load-bearing buffer stage from the time and frequency domains. Sixteen males with flatfoot and sixteen males with normal foot were recruited to walk on the plantar pressure test system at the same step speed, and the vertical ground reaction force data were collected from the heel contact stage and the arch support stage. The differences in the frequency domain of the ground reaction force between the flatfoot and the normal foot in the two stages were analyzed according to the basic mechanical characteristics and the continuous wavelet transform. Independent sample *t*-test was used to compare the baseline data of subjects, and the differences in foot impact force characteristics at different stages of foot type and weight-bearing cushion phase were compared by two-factor repeated measures Analysis of Variance. 1) In terms of basic mechanical characteristics, In both groups for flatfoot patients and normal foot people, the peak ground reaction force was higher in the arch support stage compared to the heel contact stage ($P_{flatfoot}<0.001$, $P_{normal\ foot}<0.001$), and the load rate of force change was smaller in this stage ($P_{flatfoot}<0.001$, $P_{normal\ foot}<0.001$). However, no differences in peak ground reaction force and time of occurrence were found between flatfoot and normal foot in the two stages ($p > 0.05$), in the arch support stage, the force change load rate of patients with flatfoot was lower than that of normal foot people ($p = 0.021$). 2) The results of time and frequency domain characteristics showed that during the heel contact and the arch support stage, no significant differences in the maximum signal power as well as the corresponding time and the frequency of the main impact force between the normal foot and the flatfoot were found. In both flatfoot and normal foot types, compared with the foot heel contact stage, the maximum signal power in the arch support stage was higher ($P_{flatfoot} < 0.001$, $P_{normal\ foot}<0.001$), and the corresponding impact frequency was

smaller ($P_{flatfoot}$ = 0.002, $P_{normal\ foot}$ = 0.004). Once the step speed was controlled, the flatfoot patients only showed a smaller impact force load rate in the arch support stage, which may be related to their lower arch rigidity. The characteristics of the impact force in different stages of walking support period were different in time and frequency domain, suggesting that there may be differences in the function of various parts of the foot.

KEYWORDS

gait, flatfoot, impact force, continuous wavelet transform, biomechanics

1 Introduction

Flatfoot is a foot deformity caused by the collapse of the arch [1,2]. As an important structure of the foot, the arch performs an elastic cushioning function through changes in height during movement, therefore, the collapse of the arch is considered to be an important factor that increases the risk of lower limb injury [1,2,3]. In childhood, the foot structure tends to have a flattened appearance, mostly due to incomplete foot development [4], while in elderly, the decline in foot function due to aging makes the foot muscles and ligaments unable to maintain the shape of the arch, which results in flatfoot [5]. It is believed that abnormal foot structure can affect the walking stability of the elderly and become a potential risk for falls [6]. During standing, patients with flatfoot have a higher magnitude and speed of plantar center-of-pressure trajectory sway than normal foot people and exhibit poorer static postural control [7,8], and Harrison's team showed that as the severity of flatfoot deformity increases, static postural stability decreases [7]. When walking, patients with flatfoot had higher impulse in the heel and arch regions compared to normal foot people [2], and flatfoot exhibited excessive pronation of the foot with lower center of pressure excursion index (CPEI) when pedaling and extending [9]. While running, patients with flatfoot exhibited greater range of motion in the hindfoot, midfoot, and forefoot than normal foot people, suggesting a greater risk of sports injury [10].

Plantar mechanics is an important reference for analyzing the biomechanical characteristics of flatfoot patients. In a majority of movements, the foot is the only motor organ in contact with the ground. During this time, the human body obtains external information in the form of mechanics through plantar proprioception [11,12]. With the gradual study of impact force, researchers found that the study of impact force cannot be limited to the basic characteristics of force. Nigg's team proposed that the response mechanism of human motion system to impact force should include two angles: time domain and frequency domain. Larger impact loads, impact frequencies, or soft tissue resonances may increase the risk of lower extremity injury [13,14], meanwhile, the basic mechanical indicators used to quantify the impact characteristics may not truly characterize the signals generated by foot-ground collisions [15]. Therefore, impact forces should be studied as input signals with amplitude, time and frequency domains. In the process of walking, the heel and foot arch perform passive cushioning to absorb and transfer the load [16], due to arch collapse and heel exostosis, the cause of sports injuries in flatfoot is mostly thought to be excessive local impact loading of the foot during the load-bearing buffer phase. However, it is still unknown whether changes in foot structure affect the frequency domain characteristics of plantar impact forces in patients with flatfoot and whether this change is responsible for the lower extremity injury in flatfoot. Moreover, structural variations in different parts of the foot lead to their functional differentiation, which may affect the frequency of impact forces corresponding to each part of the foot, therefore, the existence of time-frequency domain variation in different touchdown phases of the foot during the walking support period may also be a new direction to explore the dynamic function of the foot. In summary, combining basic mechanical methods and time-frequency domain analysis to investigate the plantar impact characteristics during walking not only can investigate the possible injury mechanisms in patients with flat feet, but also provides a new research method to assess dynamic foot function.

Continuous Wavelet Transform (CWT) is a time-frequency domain analysis of signals by specific wavelet functions according to different scaling scales. Compared with the Fourier Transform (FT), the CWT enhances the frequency domain analysis of signals at different times. It was found that CWT had better sensitivity compared to traditional median frequency (MF) and mean power frequency (MPF) when assessing muscle fatigue status [17]. After CWT analysis, the research revealed that for visual, proprioceptive interference and foot type differences all affected the frequency domain characteristics of the standing plantar center of pressure trajectory [7,18]. The CWT analysis of impact force also showed that rearfoot pattern runners had higher impact force frequencies during the running support period compared to non-rearfoot pattern runners [15]. The above researches all indicated that CWT has beneficial applications in biomechanical fields such as surface myoelectric signal processing and nonlinear analysis of mechanical signals. In this paper, we combine basic mechanical characteristics and CWT method to compare the load magnitude and time-frequency domain characteristics of vertical ground reaction forces during buffering phase of the

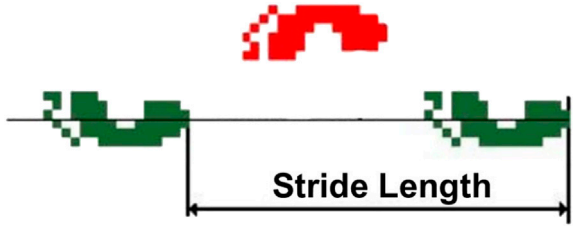

FIGURE 1
Stride length definition.

walking support period between normal foot and flatfoot. The aims of the paper is to investigate the effects of the weight-bearing cushion phase and foot type differences on plantar impact characteristics during the walking support period. The hypotheses of this study are as follows: 1) In the buffering stage of walking, the plantar impact load of patients with flatfoot is larger than that of normal foot, and the frequency of main impact force may be different from that of normal foot; 2) The weight-bearing cushion dominated by different parts of the foot during the heel touch phase and arch support phase may lead to differences in the time-domain and frequency-domain characteristics of plantar forces.

2 Materials and methods

2.1 Participants

Thirty-two male subjects were recruited. PASS (Version 15, NCSS, United States) was used for sample size estimation, the minimum effect size was set at 0.90, and the significance level was set at 0.05. When the correlation between different buffer stages within a group was 0.5 and the difference in foot type between groups was 50%, the minimum sample size determined was 22 (Tests for Two Groups of Pre-Post Scores). Combining individual differences in subjects and potential sample attrition, the total sample size was finally determined to be 32.

The foot type was distinguished by foot arch index in 3D foot scanner (3DPODOMED, Beijing Haidemi Investment Co., Ltd. accuracy ± 1.0 mm). Among them, 16 were normal foot (arch index 0.21-0.26) [19] and 16 were flatfoot (arch index >0.30) [19]. The subjects were required to have normal exercise ability (Move freely without pain or discomfort.), no lower limb injury in recent 3 months and no history of lower limb surgery. All the recruited subjects were familiar with this experimental procedure, and volunteered to participate in the experimental test. This study was endorsed by the University Ethics Committee of Hebei Normal University (No. 2022LLSC026).

2.2 Data collection

Each subject walked barefoot on a 2 m high frequency plantar pressure plate (RSscan International Company of Belgium, collecting frequency 120 Hz, sensor density 4/cm^2) for 5 times to control the walking speed and record the plantar pressure data of the dominant lateral foot closest to 1.30 m/s. The walking speed was calculated by dividing the stride length by the stride time, while the stride length and stride time were the distance and time interval between the heel touching the pressure plate twice, respectively (Figure 1). Dominant lateral foot was defined as the preferred leg for a ball kicking [20].

2.3 Data processing

2.3.1 Load-bearing buffer stage division

The time of the walking support period of each subject was normalized from 0% to 100%, and the ground reaction force during the walking support period was normalized as a percentage of the gravitational force applied to each subject, the ground reaction force diagram during the walking support period was drawn (Figure 2). In this study, we mainly analyzed the characteristics of ground impact force during the weight-bearing cushion phase (red line in Figure 2) so that we could investigate whether heel exostosis and arch collapse in patients with flatfoot would affect the foot cushioning function, and the mechanical characteristics of different parts of the foot during cushioning. As shown in Figure 1, point A is the moment of forefoot touchdown [21,22], and from the hindfoot touchdown to point A is the heel contact stage; point B is the first peak of ground reaction force [21,23], and from point A to point B is the arch support stage.

2.3.2 Basic mechanical characteristics of plantar impact

The peak value of ground reaction force, the time of peak occurrence and the load rate of force were calculated for different foot types during the heel touch phase and arch support phase, respectively, where the load rate of force was calculated by the average slope of the curve, as follows [24,25]:

$$k = f'(x)dx$$

Where k is the average slope of the curve and $f'(x)dx$ is the differential of the curve function expression.

2.3.3 Time-frequency domain characteristics of plantar impact forces

The continuous wavelet transform calculates the inner product between the wavelet basis function and the original signal at different scaling scales through time translation of the wavelet basis function, and then adjusts the time resolution and frequency resolution. Among them, the integral of the wavelet basis function

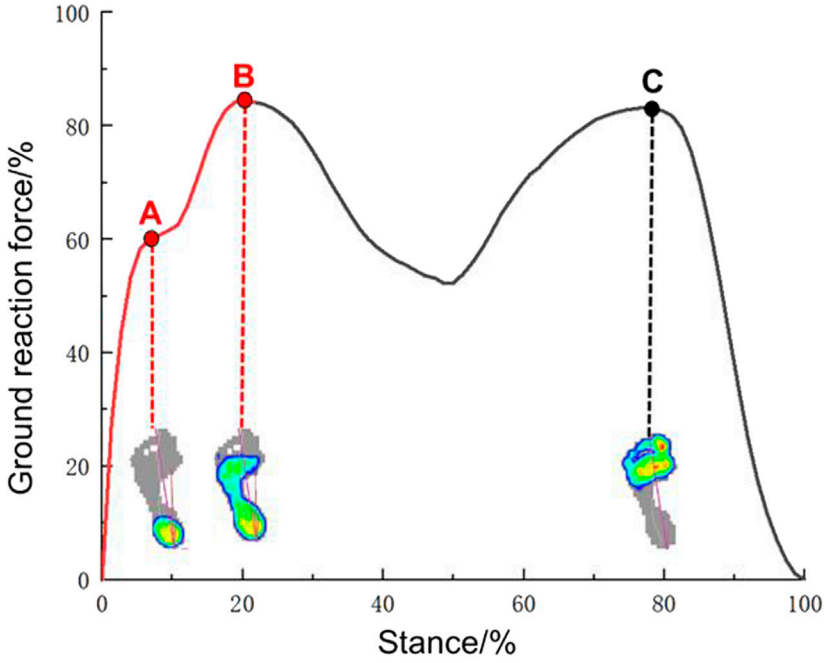

FIGURE 2
Load-bearing buffer stage division.

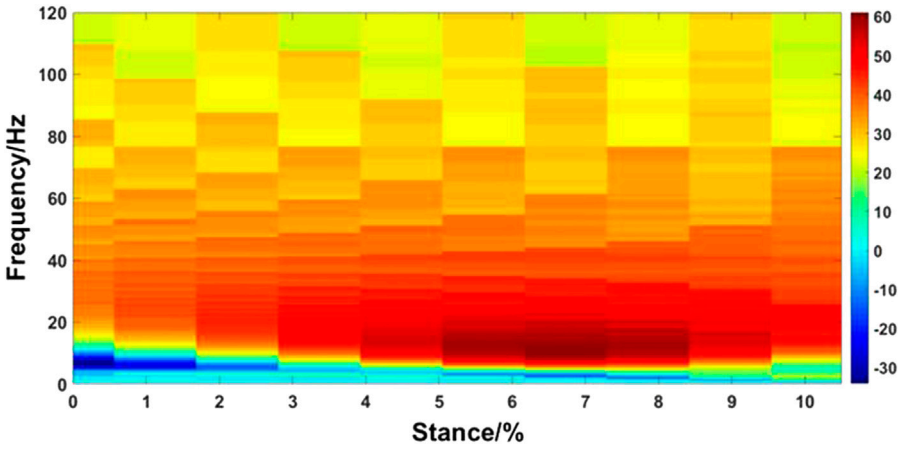

FIGURE 3
Analysis diagram of ground reaction in time and frequency domain. (Note: The color bar represents the signal power. As shown in Figure 2, the main frequency of the analytical signal is between 5–25 Hz, and the period of occurrence is in the first 5%–9% of the support period).

of the selected mother wavelet is close to 0 in its function interval, and its waveform should be similar to that of the original signal as far as possible to ensure the accuracy of the analytical results. Different scaling scales will change the center frequency of the mother wavelet. A larger scale means a lower central frequency of the mother wavelet and a narrower bandwidth in the frequency domain, so the resolution of the frequency is higher. Similarly, a smaller scale means a wider bandwidth in the frequency domain, resulting in a higher temporal resolution.

According to previous studies [15], the morlet wavelet function was selected to perform a continuous wavelet transform (Matlab 2017a, Mathwork, United States) with a scale of 1–200 (resolution frequency of 0.6-120 Hz) and a step size of 0.5 on the ground reaction force normalized to the heel touch stage and the arch support stage

TABLE 1 Baseline data of subjects (n = 32, $\bar{x} \pm s$).

	Normal foot	Flatfoot	p-value
Age (year)	25.50 ± 1.15	24.63 ± 1.54	0.080
Height (cm)	177.25 ± 2.67	177.75 ± 3.61	0.659
Weight (kg)	72.63 ± 8.21	73.88 ± 10.39	0.708
Walk speed (m/s)	1.32 ± 0.12	1.30 ± 0.09	0.683

respectively, and the time-frequency domain analysis cloud of the ground reaction force is shown as Figure 3. The indicators of this study include maximum signal power, the time of maximum signal power and the frequency of maximum signal power to reflect the time-frequency domain characteristics of the impact force.

2.4 Statistics

The mean and standard deviation for each variable were calculated. Normal distribution check of variables by Shapiro–Wilk test. Independent samples t-test was used to compare the baseline information of normal and flatfoot, and two-factor repeated measures ANOVA (Analysis of Variance) was used to compare the plantar impact characteristics of two foot types at different stages of walking support period, the Bonferroni method was used for simple effect comparisons (SPSS 25.0, IBM, United States), with the significance level α was set as 0.05.

3 Results

3.1 Comparison of subject baseline information

As Table 1 shows, there was no significant difference in age, height and weight between the flatfoot group and the normal foot group ($p > 0.05$), while no difference was found in the walking speed between the two groups ($p > 0.05$). The subjects were similar in basic condition and the experimental conditions were well controlled.

3.2 Comparison of basic mechanical characteristics of plantar impact

For the peak ground reaction force, only different buffer stages had main effect (Table 2), compared with the heel contact stage, the peak ground reaction force of both flatfoot and normal foot in the arch support stage was higher ($P_{flatfoot}<0.001$, $P_{normal\ foot}<0.001$). However, there was no significant difference in the peak ground reaction force between the flatfoot and the normal foot in the two stages ($P_{heel\ contact} = 0.793$, $P_{arch\ support} = 0.453$).

The buffer stages main effects also appeared in the time of the peak force and load rate of force change. However, no significant differences were found between foot types in the above two indicators (Table 2). Specifically, the load rate of force change was lower in both flatfoot and normal foot during the arch support stage than the heel contact stage ($P_{flatfoot}<0.001$, $P_{normal\ foot}<0.001$). In addition, in the arch support stage, the force change load rate of patients with flatfoot was lower than that of normal foot people ($p = 0.021$).

3.3 Comparison of time and frequency domain characteristics of plantar impact force

The main effect for stage was observed in both maximum signal power and the time of maximum signal power, no significant differences were found in the above indicators for foot type (Table 3). Moreover, for the maximum signal power, compared with the heel contact stage, the maximum signal power of the flatfoot and normal foot of foot arch support stage was higher ($P_{flatfoot}<0.001$, $P_{normal\ foot}<0.001$).

There was a main effect for stage in the frequency of maximum signal power, neither foot type main effect nor interaction effect were found in this indicator (Table 3). For both of the flatfoot and normal foot, the frequency of maximum signal power in arch support stage was lower than that of the heel contact stage ($P_{flatfoot} = 0.002$, $P_{normal\ foot} = 0.004$).

TABLE 2 Basic mechanical characteristics of plantar impact of different foot types.

	Heel contact stage		Arch support stage		p-value		
	Normal foot	Flatfoot	Normal foot	Flatfoot	Interaction effect	Main effect for stage	Main effect for foot type
Peak ground reaction force (%)	56.36 ± 18.19	54.33 ± 24.58	92.06 ± 14.53*a*	85.45 ± 31.57*a*	0.332	**<0.001**	0.586
The time of the peak force (%)	8.03 ± 1.83	9.06 ± 2.45	21.31 ± 5.14*a*	23.17 ± 3.33*a*	0.676	**<0.001**	0.053
Load rate of force change	5.49 ± 1.50	5.85 ± 3.30	2.75 ± 0.46*a*	2.27 ± 0.65 *ab*	0.346	**<0.001**	0.899

(Note: *a* indicates that there is a significant difference between heel contact stage and arch support stage; *b* indicates that there is a significant difference between normal foot people and flatfoot patients).
The meaning of the bold values is $p < 0.05$.

TABLE 3 Statistical results of plantar impact force in time-frequency domain.

	Heel contact stage		Arch support stage		*p*-value		
	Normal foot	Flatfoot	Normal foot	Flatfoot	Interaction effect	Main effect for stage	Main effect for foot type
Maximum signal power	63.86 ± 24.25	59.15 ± 25.52	140.89 ± 34.99*a*	143.28 ± 64.21*a*	0.650	**<0.001**	0.924
The time of maximum signal power (%)	5.77 ± 1.08	6.43 ± 1.64	16.42 ± 2.86*a*	17.45 ± 2.19*a*	0.721	**<0.001**	0.122
The frequency of maximum signal power (Hz)	8.01 ± 1.90	7.64 ± 1.80	5.33 ± 2.06*a*	4.64 ± 1.76*a*	0.797	**<0.001**	0.056

(Note: *a* indicates that there is a significant difference between the heel contact stage and arch support stage).
The meaning of the bold values is $p < 0.05$.

4 Discussion

4.1 Analysis of basic mechanical characteristics of plantar impact

Walking is one of the basic movements of the human body, the foot, as the end link in direct contact with the ground during walking, has a complex structure that allows the human body to accomplish weight-bearing cushioning and stretching functions. Due to the slower speed during walking, all parts of the foot can fully contact the ground; [12,26,27]. In the load-bearing buffering stage, the heel first touches the ground, at which time the strike between the heel and the ground produces a peak force, as shown at point A in Figure 1, which is known as the "heel strike transient" [21,23], and then the foot rolls forward around the heel axis, the midfoot gradually touches the ground, when the arch plays a dominant role in cushioning [23]. At this time, the human body adjusts the foot posture according to the characteristics of the ground reaction force, while viscoelastic structures such as foot fascia stores elastic potential energy [12,28]. The results of this study showed that the slope of the ground reaction of the two types of foot in the arch support stage was lower than that in the heel contact stage ($p < 0.05$). In the heel contact stage, although the human body is in the double-support phase while the contralateral limb bears part of the impact load, the load rate of change in the vertical ground reaction force is still higher than that in the arch support phase, which indicates the arch structure of the foot effectively reduces the impact of ground reaction force, reflecting the elastic cushioning function of the foot.

In the arch support stage, our study found that the force change load rate of flatfoot patients was lower than that of normal foot people ($p < 0.05$), suggesting that the arch of the normal foot may exhibit greater rigidity in response to impact loading, which is consistent with the previous findings of our group [29]. A finite-element analysis of flatfoot [30]. found that lower plantar fascia rigidity and inadequate strength of the peroneus longus affects the ability to maintain the arch of the foot, the result also explains the findings of our study. From these results, it can be concluded that for patients with flatfoot, the lower arch rigidity requires the selection of orthotic insoles with better foot support for functional foot assistance. However, no significant differences were found between normal and flatfoot in the peak ground reaction force and the time of peak ground reaction during both heel contact and arch support stage. A study of cadaveric foot [31] showed that sustained impact loading increased the flatness of flatfoot, this finding suggests that flatfoot is less adaptive and prolonged exercise may further amplify the differences in foot function between flatfoot patients and normal foot people. Since the physical differences between the subjects in this study were small and the step speed was controlled during the test, the foot type differences have less effect on the peak plantar impact force, therefore, it can be speculated that the main factors affecting the magnitude and corresponding time of ground reaction force during the walking buffer period may be the step speed and the effective mass at the time of impact.

4.2 Time-frequency domain characteristics analysis of plantar impact force

In the field of sports medicine, earlier studies mostly considered that the main cause of sports injuries was the accumulation of load or excessive impact, and the biomechanical studies of basic sports such as walking, running and jumping mostly took the time domain perspective as the entry point. With the in-depth study of frequency domain, scholars found that human bones (200–900 Hz) [13,32] and soft tissues (10–60 Hz) [33] have corresponding inherent frequencies respectively, while the impact force frequency domain during running is around 10–20 Hz [15,34], which is closer to the inherent frequency of soft tissues, making the impact force frequency and the inherent frequency of soft tissues resonate and lead to soft tissue injury. This view provides a new perspective on the mechanism of sports injury.

During walking, all parts of the foot are in full contact with the ground, which makes the foot function more sufficiently embodied. In this study, the ground reaction force in the load-bearing cushion phase during the walking support period was

considered as a set of input signals with time-frequency domain characteristics, however, in the time-frequency domain perspective, our study did not find significant differences in the maximum signal power, corresponding time and impact force frequency between flatfoot patients and normal foot people during the load-bearing cushion phase, which suggests that after controlling for step speed, arch collapse has little effect on the frequency domain characteristics of plantar impact during low-intensity and short-term exercise, while significant changes in plantar impact characteristics may need to be further reflected by high-intensity or prolonged exercise.

This study showed that the maximum signal power of the arch support stage is higher than that of the heel contact stage, indicating that the plantar impact is mainly concentrated in the arch support stage. Meanwhile, the arch support stage exhibited a lower impact force frequency compared to that of the heel contact stage. Some scholars believe that the musculoskeletal system of the human body plays the role of a "low-pass filter" during movement, and its purpose is to ensure the stability of the head during movement so that the brain can accurately process visual and vestibular inputs and improve movement stability to prevent falls [35,36], while the angle of the joint, the degree of muscle activation, and the skeletal muscle composition all change the cut-off frequency of the "filter" [14,36,37]. The results of this study support this argument, even in different parts of the foot, structural differences can also make a change in the frequency of the impact force on each part of the foot. Shorten's team found that high-frequency impact force was mainly present in the initial landing phase of running, while low-frequency impact force accounted for a gradual increase in the middle of running support [38]. Although the movement studied in our paper was walking, due to the similarity of movement patterns, the results of both this research and Shorten's team suggest that there may be differences in impact frequencies in various parts of the foot due to structural differences. In the heel contact stage, the human body responds to ground impact with the ankle-hindfoot as the dominant, the heel bone is subjected to larger rigid impact resulting in a higher force change load rate and impact force frequency. While in the arch support phase, the weight-bearing cushion of the foot is dominated by the arch, and the mid-tarsal joint and plantar fascia sufficiently carry out elastic cushioning, thus showing smaller force change load rate and impact force frequency. The above differences indicate that frequency domain characteristics can be a new perspective for evaluating foot function.

In this study, the time-frequency domain analysis provided a new perspective on the plantar impact characteristics during the weight-bearing cushioning phase of the walking support period, but there are still some limitations in our study, the present research only investigated the time-frequency domain characteristics of impact force under low-intensity exercise like walking in male flatfoot patients, while the motion characteristics under high-load impact such as running and jumping in female and different age groups of patients with flatfoot also need some data support, and it is suggested that future studies should be supplemented in this aspect.

5 Conclusion

After controlling the gait speed, it is hard to find the difference in impact force frequency between flatfoot patients and normal foot people during the weight-bearing cushioning phase, the smaller impact force change load rate in the arch support stage of flatfoot patients may due to their lower arch rigidity, the better foot support footwear is recommended for flatfoot patients. Variations in foot function in different parts of the foot may cause different frequency domain characteristics during the weight-bearing cushioning phase.

It is recommended that future research on the plantar mechanics of flatfoot should extend the sample size and increase the study of injury mechanisms in people of different genders and ages during prolonged or high-intensity exercise.

Data availability statement

The raw data supporting the conclusion of this article will be made available by the authors, without undue reservation.

Ethics statement

The studies involving human participants were reviewed and approved by the University Ethics Committee of Hebei Normal University (No. 102772021RT034). The patients/participants provided their written informed consent to participate in this study.

Author contributions

XB and HH were in charge of designing the experiment. XB wrote the manuscript. PL, YL, AK, and JL revised the manuscript. All authors have read and approved the final manuscript.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Toward improved understanding of foot shape, foot posture, and foot biomechanics during running: A narrative review

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The current narrative review has explored known associations between foot shape, foot posture, and foot conditions during running. The artificial intelligence was found to be a useful metric of foot posture but was less useful in developing and obese individuals. Care should be taken when using the foot posture index to associate pronation with injury risk, and the Achilles tendon and longitudinal arch angles are required to elucidate the risk. The statistical shape modeling (SSM) may derive learnt information from population-based inference and fill in missing data from personalized information. Bone shapes and tissue morphology have been associated with pathology, gender, age, and height and may develop rapid population-specific foot classifiers. Based on this review, future studies are suggested for 1) tracking the internal multi-segmental foot motion and mapping the biplanar 2D motion to 3D shape motion using the SSM; 2) implementing multivariate machine learning or convolutional neural network to address nonlinear correlations in foot mechanics with shape or posture; 3) standardizing wearable data for rapid prediction of instant mechanics, load accumulation, injury risks and adaptation in foot tissue and bones, and correlation with shapes; 4) analyzing dynamic shape and posture *via* marker-less and real-time techniques under real-life scenarios for precise evaluation of clinical foot conditions and performance-fit footwear development.

KEYWORDS

foot shape, foot posture, statistical shape modeling, principal component analysis, foot biomechanics

Highlights

- The artificial intelligence is a useful metric of foot posture, which can be measured quickly and associated with motion and joint pain in adults, but may be less useful in special populations such as developing and obese individuals.
- The foot posture index that is used for classifying feet as pronated, neutral, and supinated is a useful approach under both static and dynamic conditions. However, care must be taken to associate pronation with injury risk, and factors of the Achilles tendon and longitudinal arch angles may be required for elucidating the risks.
- Dynamic tracking of foot shape, posture, and internal multi-segmental motion should consider current fluoroscopy, machine learning, statistical shape modeling, wearable, and marker-less techniques.

Introduction

Changes of shape and posture in the human foot have been informed through the evolutionary process (Bennett et al., 2009; Lieberman, 2012; Harcourt-Smith et al., 2015). Foot shape and functions are closely related and especially adapted to bipedalism. While examining the foot from a functional perspective, the posterior calcaneus contributes to balance support and impact absorption, the dome-like arch stiffens the foot during weight-bearing and returns energy during push-off in gait, the metatarsals expand the anterior pressure support, and the toes (particularly the hallux) facilitate pushing-off at the end of stance during locomotion (Morton, 1924; Hicks, 1953; Hicks, 1954; Ker et al., 1987; Ward et al., 2011; Fernández et al., 2018; Venkadesan et al., 2020).

While considering the measurement of foot functions, the classification of the foot–ankle complex has been presented using several popular metrics. A commonly employed model in clinics was based on the "Rootian theory" (Root model), also known as the "subtalar joint neutral theory" (Root et al., 1977). This model uses static measurement of the subtalar joint neutral position to predict dynamic functions and prescribe orthotics for treatment. However, this model has been challenged recently for poor (low) correlation with dynamic functions (Jarvis et al., 2017) and concerns of reliability and validity (Harradine et al., 2018). The technique of classifying foot posture types has been reported using visual inspection, anthropometric measurement, footprint analysis, and radiographic assessment for rearfoot, midfoot, and forefoot (Razeghi and Batt, 2002). These include measurement of the arch height, longitudinal arch angle, navicular drop and drift, artificial intelligence and arch angle, and radiographic evaluation of calcaneal inclination angle, height–length ratio, calcaneal–first metatarsal angle, and rearfoot–forefoot angle. The foot posture index (FPI) is another popularly employed metric to define pronated, neutral, or supinated feet *via* anatomical palpation and structural observation (Redmond et al., 2006). In terms of dynamic conditions, a recent minimal markerset model was proposed for navicular position measurement and validated for relating foot postures and functions with accurate intraday reliability (Eichelberger et al., 2018).

In the current literature, several excellent studies have reviewed the relationship of foot shape, foot posture, and foot biomechanical function. Shoe-wearing habits, pathology, or external factors have also played contributing roles in foot shape, posture, muscle and tendon morphology, and bone alignment (Cavanagh et al., 1997; Johnson et al., 2015; Xiang et al., 2018; Garofolini and Taylor, 2019). Specifically, a 10-week transition into running with minimalist shoes found an increased cross-sectional area of abductor hallucis by 10.6% (Johnson et al., 2015). This has confirmed that tissue morphology is related with shoe-wearing habits of minimalist shoes, motion control shoes, and neutral shoes (Zhang X. et al., 2018). Decreased hallux abductus angle (measured from X-ray images) and hallux angle (measured with footprints) have been reported with centrally shifted plantar pressure and reduced medial metatarsals stress followed by a 12-week minimalist shoe intervention for mild hallux valgus (Xiang et al., 2018; Xiang et al., 2022b).

In particular, the foot type and foot disorders are strongly associated with the foot arch difference, classified as pes cavus (high arch), pes rectus (normal arch), and pes planus (flat arch) (Hillstrom et al., 2013; Mootanah et al., 2013). Pes planus feet have been associated with hammer toes and overlapping toes, while pes rectus and pes cavus have not been associated with any foot disorders (Hagedorn et al., 2013). Furthermore, decreased thickness and area in the intrinsic muscles (such as abductor hallucis, flexor hallucis brevis, and peroneus longus and brevis), plantar fascia, and Achilles tendon, while increased thickness and area in the extrinsic muscles (flexor digitorum and flexor hallucis longus) have been found in pes planus, which may implicate the compensatory adaptation for the altered foot structure (Angin et al., 2014, Angin et al., 2018; Crofts et al., 2014; Murley et al., 2014). However, these measurements or analyses were conducted under static conditions.

During walking (dynamic conditions), cavus (high arch) feet presented increased frontal and transverse motion in the rearfoot, while planus (flat arch) feet showed a reduced frontal range of motion in the midfoot (Buldt et al., 2015a). The foot posture index (FPI) showed a stronger correlation with intersegmental kinematics than did the other measurements of the artificial intelligence, navicular height, and dorsal arch height (Buldt et al., 2015b). Additionally, planus feet showed greater activation of the tibialis anterior but less activation of peroneus longus during the initial contact of stance and increased tibialis posterior but decreased peroneus longus activities during midstance and the push-off phase (Murley et al., 2009b). It has been further reported that increased foot inversion and

muscle activation were present in pronated feet with less evertor activation (Murley et al., 2009a).

Alteration of shape or posture in the foot, functioning as the primary interface with the external surroundings, may lead to the realignment of the kinetic chain. Foot pronation often showed an everted rearfoot and arch drop in the foot–ankle complex, further inducing internally rotated tibia, proven with a continuous vector coding analysis technique (Rodrigues et al., 2015), and mobile patellar and anterior knee pain (Clifford et al., 2020; Yu et al., 2021) in the knee complex. During the changes in the kinetic chain, the pronated foot posture was further reported to be associated with medial tibia stress injury (Neal et al., 2014), increased subtalar motion, leg stiffness, tibial shock (Hollander et al., 2019), and patellofemoral pain (Selfe et al., 2016) during running. While the posture alteration is not strongly related to lower back pain, apart from females in whom pronated foot showed back pain (Menz et al., 2013). Pronated feet have been strongly associated with hallux valgus and overlapping toes, while supinated feet have shown less association (Hagedorn et al., 2013).

In terms of foot postures and shape in the foot segments, particularly when analyzing the functions of the forefoot *in vivo*, a longer first metatarsal size, rounder first metatarsal head, and greater first metatarsophalangeal joint angle were associated with hallux valgus deformity (Nix et al., 2012). Computational modeling of the foot (Morales-Orcajo et al., 2016) revealed a higher concentrated von Mises stress at the metatarsals and higher contact pressure in the first metatarsophalangeal joint (Zhang et al., 2018c). Furthermore, the diabetic foot presented toe deformation with focalized pressure at the hallux (Lu et al., 2015). Habitually, barefoot populations with increased hallux spacing have been associated with active gripping function (Lambrinudi, 1932; Wallden, 2016), which expands the supporting area in the forefoot (Mei et al., 2015a; Mei et al., 2015b; Shu et al., 2015; Wang et al., 2016) and higher medial longitudinal arch in the barefoot children's cohorts (Hollander et al., 2017). A nontypical foot shape, such as the bound foot in Chinese women, presents dislocated toes, extreme high arch, limited ankle range of motion, and concentrated loading in the heel (Gu et al., 2015; Zhang et al., 2018b).

Recently, there has been a focus on distance running activities (which is in contrast to static measures or walking) and population-based modeling using large data sets and statistical shape methods (Fernandez et al., 2016; Fernandez et al., 2019). In this study, we first aimed to review and discuss the changes in foot morphology, shape, and posture during dynamic activities, using running as an example. Second, we examined the statistical shape modeling (SSM) of the foot using population-based approaches and reviewed the association between shape, posture, and biomechanics to understand potential injury. Lastly, we summed up the current advances and techniques in analyzing the foot shape, posture, and biomechanical functions and provided several perspective suggestions for consideration in the foot–ankle health-related research.

Foot morphology, shape, and posture in running activities

Evaluation of foot morphology, shape, and posture

Changes in tissue morphology, bone shape, and posture of the foot influence the biomechanics of the lower extremity. For example, foot types (mainly, normal, planus, and cavus feet) (Buldt et al., 2015a), foot posture (mainly, normal, pronated, and supinated feet) (Hollander et al., 2019), toe morphology (Mei et al., 2015a), hallux valgus (Hannah et al., 2016), and manipulated forefoot shapes (abducted hallux *versus* adducted hallux) (Mei et al., 2016; Xiang et al., 2020a; Xiang et al., 2020b) have been previously reported in the literature. Pathological conditions, such as diabetic-related foot deformities (Guiotto et al., 2013; Lu et al., 2015) can also influence the biomechanics. Figure 1 outlines the most common popular shape metrics (Figure 1) and postures in the foot (Figure 1 and Figure 2) with highlighted regions of variations in pronation (blue) and supination (red) when compared to those of the neutral foot posture. The highlights in pronated and supinated foot postures are modes generated from the principal component analysis (PCA) in the statistical shape modeling (SSM) using an open-source Musculoskeletal Atlas Project (GIAS2 package) developed at the Auckland Bioengineering Institute (Zhang et al., 2014). The SSM typically refers to the techniques of describing the characteristics of deformable objects with different shape features, presenting mean shape (appearance) and key variations in these object groups (Sarkalkan et al., 2014; Mei et al., 2021a).

The current measurement of foot morphology, shape, and posture using 2D footprints, anthropometric measurements, anatomical palpation, 3D surface scanning, ultrasound imaging, computed tomography (CT), or magnetic resonance imaging (MRI) are mainly under static (non-weight or weight-bearing) conditions (Razeghi and Batt, 2002; Redmond et al., 2006; Crofts et al., 2014). However, the foot may present different postures, shapes, and morphologies during dynamic activities (short and long term). Prolonged repetitive impact on bones and soft tissue can alter the shape, biomechanical response, and adaptation (Garofolini and Taylor, 2019). In particular, increased muscle volume and cross-sectional area of foot muscles (abductor halluces, flexor digitorum brevis, and abductor digiti minimi) and bone density (calcaneus) were concluded in a recent review study (Garofolini and Taylor, 2019), which may rely on training volume and experience.

Long-distance running, or endurance running, is a key evolutionary skill presented by bipeds in addition to walking

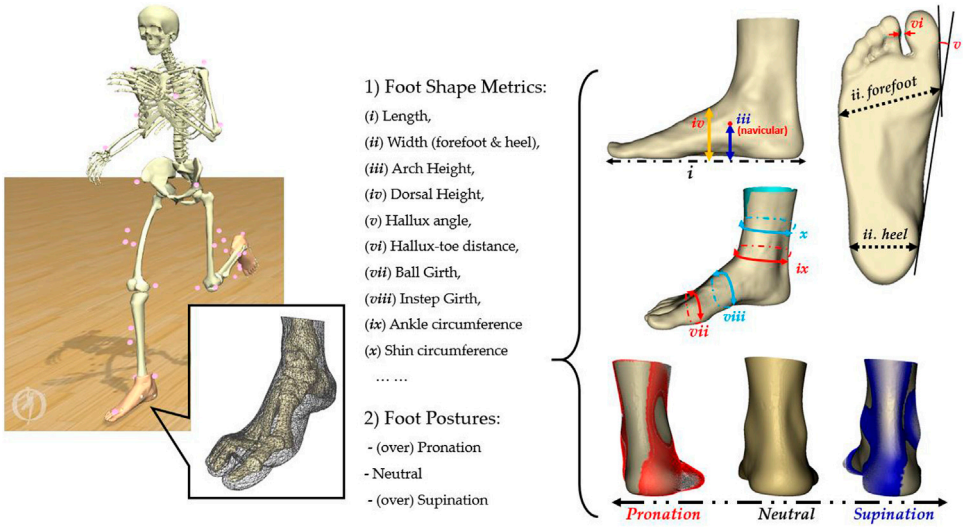

FIGURE 1
An illustration of popular foot shape metrics and foot postures with highlighted variation of pronation (red regions) and supination (blue regions) from the posterior view.

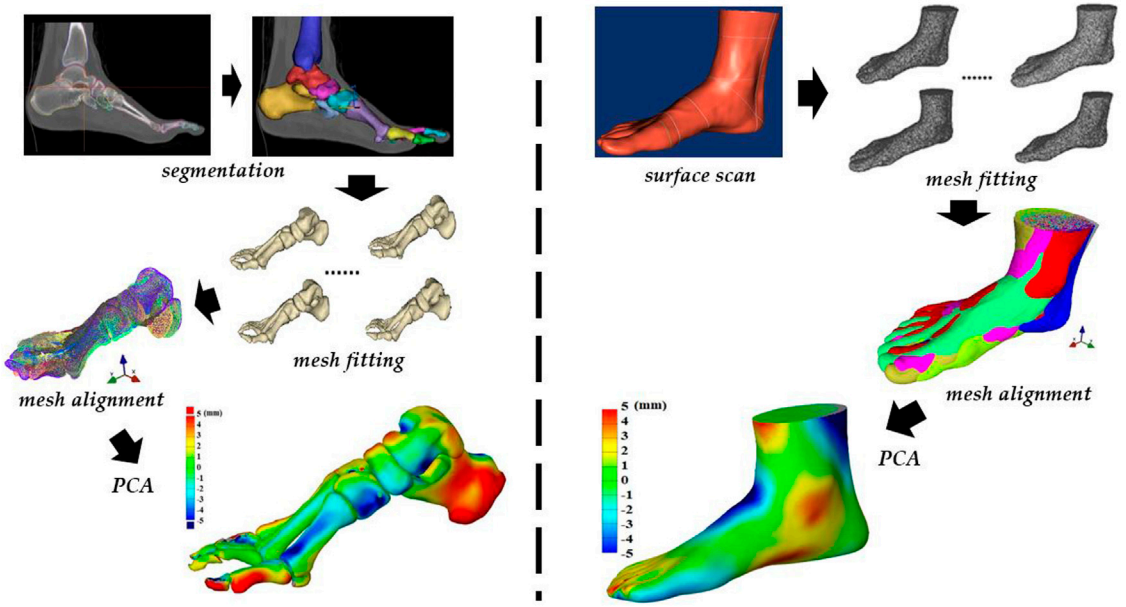

FIGURE 2
An outline of workflow in the statistical shape modeling (SSM) of foot bones (left) and surface (right) shapes. Steps of segmentation (bone) or surface scan (surface), mesh fitting, mesh alignment, and PCA.

(Bramble and Lieberman, 2004). Long-distance running has gained global popularity with increased marathon participation, particularly in amateurs. However, the increased participation is associated with higher risks of running-related injury (Hulme et al., 2017).

The arch (either high arch or flat foot) in the midfoot region is commonly used to classify and categorize the foot posture in foot biomechanics. Specifically, it has been quantified using the 2D footprint artificial intelligence (AI) and classified as high arch (AI < 0.21), normal arch (0.21 < AI < 0.26), and low arch (AI >

0.26) (Cavanagh and Rodgers, 1987). This metric has been used during static and dynamic measurements (Scholz et al., 2017) and is sensitive to forefoot and midfoot plantar pressures during walking and running (Mei et al., 2020). The lower arch has been associated with increased hallux and medial mid-foot pressures and reduced medial forefoot pressures (Jonely et al., 2011). Increased reported ankle and knee pain have also been associated with lower foot arch (Riskowski et al., 2013), whereas the high arch is primarily associated with ankle pain only (Riskowski et al., 2013). Altered frontal and transverse rearfoot motion and reduced midfoot motion during initial contact are associated with high arch feet, while reduced midfoot motion during pre-swing is associated with low arch feet (Buldt et al., 2015a). While the foot arch is a useful measure in adults, there is a caveat that it might be a less useful measurement in developing and overweight individuals. For example, the dynamic artificial intelligence was not associated with any kinematics, kinetics, or spatiotemporal information in children except for a greater external foot rotation associated with a lower arch (Hollander et al., 2018b). The artificial intelligence is also influenced by training levels and running experience (Fascione et al., 2009) despite the biomechanical evaluation being not reported. Overweight adults (with high BMI) showed pronated and flatter feet with reduced ankle inversion–eversion motion and higher plantar loading underneath the foot (Butterworth et al., 2015), which led to uncertainty of either the change of foot structure or increased weight contributing to a greater plantar loading.

Relationship between morphology, shape, and posture with running biomechanics

Running biomechanics influences the foot structure. Following long-distance running, foot arch and dorsal height reduce over a week and may take more than a week to return to pre-run profiles and has been linked with increased plantar loadings in the medial foot (metatarsals and arch) (Fukano and Iso, 2016; Fukano et al., 2018; Mei et al., 2018). These changes could affect the perceived comfort of the running footwear and contribute to running-related injuries (Cowley and Marsden, 2013; Hollander et al., 2018a). Following a running-induced fatigue intervention, plantar pressure has been reported to redistribute in the lateral metatarsals of flat arch feet and the medial metatarsals of high arch feet (Anbarian and Esmaeili, 2016). Following long-distance running, a pronated foot posture, reduced arch height, and increase in medial plantar pressure have been reported (Nagel et al., 2008; Schlee et al., 2009; Cowley and Marsden, 2013). The FPI (Redmond et al., 2006) following long-distance running has been moderately correlated with knee and ankle joint loads using the musculoskeletal model (Mei et al., 2019). Furthermore, runners with asymptomatic overpronated feet have larger abductor hallucis and flexor digitorum brevis and longus but smaller abductor digiti minimi, and peak eversion in the rearfoot and peak supination in the forefoot (Zhang et al., 2017). While, symptomatic overpronated runners have shown a smaller cross-sectional area in the flexor digitorum longus and abductor hallucis and thinner peroneus and abductor hallucis than those asymptomatic pronated runners, implying the training of intrinsic foot muscle for the possible prevention of injuries (Zhang et al., 2019).

Associations between foot pronation and injury mechanisms

Foot types (mainly, normal, planus, and cavus feet) and postures (mainly, normal, pronated, and supinated feet) have been reported to be associated with running-related injuries (Pérez-Morcillo et al., 2019), particularly the tibia stress under extreme foot types (Barnes et al., 2008), medial tibia stress, and patellofemoral pain with foot pronation (Lun et al., 2004; Neal et al., 2014; Selfe et al., 2016). The hip joint loading was found to be increased, whereas only a moderate correlation with foot pronation and ankle and knee loadings was reported (Mei et al., 2019).

This subsection is focused on foot pronation and potential contribution to running-related injuries, taking distance running as a typical example. Overpronated feet have been implicated in developing overuse injuries, despite there being no scientific evidence that overpronated feet are associated with the diagnosis of injuries or diseases. This misconception exists possibly because altered foot postures have often been observed in people who have musculoskeletal injuries, dysfunctions of the lower limb, and lower back pain (Levinger et al., 2010; McWilliams et al., 2010; Sharma et al., 2010; Menz et al., 2013; Arnold et al., 2019).

Pronation/supination are important biomechanical functions in gait, and a certain extent of natural pronation/supinations is required as a shock absorber during the early stance phase and as a rigid lever to push forward during the terminal stance phase (Hetsroni et al., 2008). During repetitive movements such as long-distance running, however, the high volume of impact forces during the early stance phase may lead to overpronation (Mei et al., 2019) by flattening the foot with arch collapses and transferring the foot eversion into the internal rotation of the tibia. There has been a belief that this overpronation possibly leads to overuse injuries by disrupting the coupling mechanism of the lower limb alignment as the gait may be compromised and adding the additional strain of the foot/ankle complex (Subotnick, 1985). Pronation is also a passive force that occurs within the initial heel strike during walking and running. That is, there would be less muscular control when the foot is overpronated, resulting in a lack of normal distribution of

excessive force and instability of the foot/ankle (Subotnick, 1985). This patho-mechanical alteration of foot posture, therefore, has been proposed to cause foot diseases such as plantar fasciitis (Golightly et al., 2014), osteoarthritis (Reilly et al., 2009; Lithgow et al., 2020), metatarsalgia (Eustace et al., 1993), and stress fractures of the lower limb (Lysholm and Wiklander, 1987).

Overpronated feet, specifically at the subtalar joint, have been proposed to develop deterioration of the lower limb joint by disrupting normal alignment with external rotation of the tibia and calcaneal inversion which are not in the normal direction (Tiberio, 1987; Hintermann and Nigg, 1998). A previous study also showed that people with medial compartment knee osteoarthritis revealed a more pronated foot than healthy controls, possibly due in part to genu varum malalignment of the knee, which causes compensatory overpronation of the pronated foot (Levinger et al., 2010). Pronation with adduction of the talus with calcaneus eversion results in a greater compressive force in the medial midfoot, and this idea has been supported by several studies, reporting that people who have been diagnosed with midfoot osteoarthritis have a more pronated foot posture (Menz et al., 2010; Arnold et al., 2019; Arnold et al., 2021; Lithgow et al., 2020). Older adults, who were diagnosed with radiographic osteoarthritis of the talonavicular joint and navicular–first cuneiform joint, presented flatter feet with greater loading of the midfoot during walking (Menz et al., 2010). This mechanism has also been supported by a cadaver study, reporting a greater compression force at the dorsal talonavicular joint simulating a flattening foot (Kitaoka et al., 1996).

However, while numerous studies have observed pronation/supination of the foot in people who have lower limb diseases, there is no scientific evidence supporting the causal relationship between the alteration of foot posture and diagnosis of injuries or diseases. A recent study on knee osteoarthritis with foot posture also put forward the question of foot postural changes leading to injuries or injuries resulting in posture changes (Al-Bayati et al., 2018). A comprehensive cohort study of running-related injuries in a large population of 1,680 runners reported that there was no significant association between anthropometric outcomes (e.g., high/low arch and rearfoot valgus) and risk factors of running-related injuries (Water et al., 1960). Similarly, a 1-year epidemiological prospective cohort study of 927 novice runners reported that foot pronation is not associated with an increased risk of running-related injuries (Nielsen et al., 2014). A systematic review study revealed a small effect between foot pronation and the risk of medial tibial stress syndrome, suggesting that foot pronation may not be directly associated with the foot injury (Neal et al., 2014). Furthermore, a more recent radiologic study reported that significant ankle kinematic changes associated with supination of the foot were not related to the diagnosis of diseases (Kim et al., 2019a; Kim et al., 2021). This study interestingly observed that novice runners, who did not show ankle kinematic changes after mid-distance barefoot running, revealed early indications of cartilage degeneration or deteriorating effects by increasing the T2 relaxation time in MRI-derived T2 maps. However, runners who showed a supinated foot type after mid-distance running did not change their T2 value on MRI. This study, therefore, suggests that supinated feet or significant ankle kinematic changes are less likely to develop foot/ankle injuries.

The idea of our understanding of the association between pronation/supination and running-related injuries is still not clear and no consensus has been reached on the foot posture with injuries (Nigg et al., 2019). Thus, to draw a conclusive result, we may require additional information such as the integrated longitudinal arch angle and Achilles tendon angle proposed for the determination of foot postures and biomechanics during walking and running (Limeres et al., 2019; Behling and Nigg, 2020). Due to multifactorial parameters being included for analysis, multivariate statistical models, such as principal component analysis (Limeres et al., 2019; Behling and Nigg, 2020), partial least square regression (Mei et al., 2020), and other nonlinear statistical models, are recommended to investigate the potential correlation.

Numerous foot imaging modalities have been adopted to capture the foot posture *via* shape which include the 3D plantar surface (Kimura et al., 2008; Thabet et al., 2014), dorsal surface (Blenkinsopp et al., 2012), and whole foot (Boppana and Anderson, 2019). Techniques of tracking *in vivo* foot motion have been previously developed and validated using makers in a biplane fluoroscopy system (Iaquinto et al., 2014), and midfoot postures have been analyzed to evaluate the longitudinal arch angle under conditions of barefoot (~127.5°), footwear (~130°), and orthoses (~131°) (Mannen et al., 2018). However, challenges of multiaxial motion in the ankle complexity of tibiotalar and subtalar joints were found (Canton et al., 2020). The development of *in vivo* bone shapes in the foot has laid the foundation of classifying foot types from 3D perspectives (Ledoux et al., 2006). Recent attempts have been made to associate static foot bone images from 3D CT with 2D biplanar video-radiography images *in vivo* (Maharaj et al., 2020) and the development of the multi-segmental foot musculoskeletal model (Malaquias et al., 2017; Maharaj et al., 2021), but the cost of obtaining subject-specific bone geometry makes this method less translatable for practical use. Population-based modeling using 'big data' may provide an alternative for rapidly creating foot geometries from limited data [such as the popularly employed statistical shape modeling of functional foot bones (Grant et al., 2020)] and establishing relationships between form and function at the population level.

Population-based modeling of shapes

Statistical shape modeling with principal component analysis

Statistical shape modeling (SSM) is a reduction technique that can be used to identify independent (orthogonal)

geometrical features of a set of similar shapes and rank them (Cootes et al., 1995). The principal component analysis (PCA) is the most popular method to reduce dimensionality and computes the mean shape and orthogonal shape variations (modes) (King and Eckersley, 2019). This technique has been used widely in the biomechanics and anatomy space (Ambellan et al., 2019; Audenaert et al., 2019; Fernandez et al., 2019; Vallabh et al., 2019; Wang and Fernandez, 2020; Yeung et al., 2020). Applications include clinical medical image analysis (Heimann and Meinzer, 2009) and surgery and design of orthopedic implants (Sarkalkan et al., 2014).

To introduce one of the SSM technique, we took musculoskeletal modeling software GIAS2 (Zhang et al., 2014) (https://pypi.org/project/gias2/) being developed at the Auckland Bioengineering Institute as an example. Figure 2 demonstrates the workflow in the SSM of the foot surface shape (right side) and bone shape (left side) being employed from our research group. The steps include shape mesh segmentation (bone shape) or surface scan (surface shape), mesh fitting and alignment, and the principal component analysis (PCA) to compute the mean shape and key modes of variation.

Population-based approaches for understanding the foot have been reported in osteoarthritis populations (Trivedi et al., 2010), for the footwear type and foot discomfort in females (Dufour et al., 2009), flat arch with ankle pain and high arch with knee pain (Riskowski et al., 2013), and hallux valgus with foot pain (Dufour et al., 2014). Recent advances in foot imaging technology have enabled easy access and increased availability of 3D foot scanning systems (Telfer and Woodburn, 2010), providing increased foot morphology data for the study of shape variations, especially in populations of different ethnicities (Mei et al., 2021b; Mei et al., 2021a). A recent study highlighted published 1.2 million foot shape data on the features of populations from North America, Europe, and Asia and reported the distribution of geometrical metrics (such as length, width, and height) between males and females (Jurca et al., 2019). Apart from the measurements from 3D scanning technologies, the PCA on the 3D shape is one common and useful approach to extract meaningful information from these data (Mei et al., 2021a).

The SSM of the calcaneus and talus bones in the rearfoot revealed a smaller size in females but less asymmetry in both genders (Audenaert et al., 2019), specifically in the length and height of calcaneus and talus articular surface (Tümer et al., 2019a). In contrast to the normal foot, the calcaneus shape presents decreased height and increased length in high arch feet and increased posterior mass in the talus of flat arch feet (Moore et al., 2019). A statistical shape model of the foot surface reported variations in arch height, ball (metatarsophalangeal joint) width, toe distance, hallux orientation (valgus–varus), and toe length (Stanković et al., 2018). A higher BMI has been related to greater ankle width, Achilles tendon size, and width. Age was associated with heel width, Achilles tendon size, and hallux orientation. Gender was linked to ankle width, Achilles tendon size, and heel width. Classifying problematic feet using surface morphology has been reported which included hallux valgus, pes planus, and pes cavus profiles (Stanković et al., 2020).

A flat talar contact surface has been associated with chronic ankle instability (Tümer et al., 2019b). This flat talar surface may be associated with adaptation to constraining footwear on the basis of comparisons with the talus shape from archaeological records (Sorrentino et al., 2020). High arch feet have a more posteriorly positioned navicular tuberosity than normal arch feet (Moore et al., 2019). In the forefoot, metatarsal shapes are associated with foot types (high arch, normal, and flat arch) and bones vary in size (Telfer et al., 2017). Specifically, the flat arch foot has a reduced cross-sectional metatarsal area, and the first and fourth metatarsal sizes are linked to gender, with females presenting smaller sizes. Furthermore, the first metatarsal could be accurately reconstructed from sparse landmarks based on the SSM (Grant et al., 2020), which has the potential for rapid development of the musculoskeletal foot models (Grant et al., 2020).

Recent techniques to correlate shape with function in the foot

One of the challenges with the SSM is creating a population of topologically consistent geometries for the principal component analysis. This problem has been addressed by using the 'free-form' deformation technique (Fernandez et al., 2004), where a generic geometry is morphed to different data ('host mesh'). This has been demonstrated for the diabetic foot (Fernandez et al., 2012) and gout foot (Dalbeth et al., 2015) and, recently, for investigating ankle pressure in barefoot runners (Kim et al., 2019b). This technique has been used to check the similarity of foot shapes (Mochimaru et al., 2000), which are combined with plantar pressures for the application of footwear design (Kim et al., 2007). Another consideration which should be noted is that the prediction of four-dimensional (3D shapes varying over time) foot shapes become plausible from integrating multidisciplinary advanced statistics, artificial intelligence (AI), depth camera, and object detection techniques (Boppana and Anderson, 2019; Boppana and Anderson, 2021). The development of this technology may become a great improvement in shape changes in real time and provide an option of considering footwear and orthotics fit from a dynamic and functional perspective.

Recently, multivariate machine learning (partial least squares) regression models were first developed to correlate key shape metrics (artificial intelligence and hallux–toe distance) with walking (R-square values of 0.763 and 0.788) and running (R-square values of 0.786 and 0.789) plantar pressures using habitually barefoot and shod populations (Mei

et al., 2020), showing a prediction accuracy of around 80%. Further sensitivity analysis reported that the forefoot shape metric (hallux–toe distance) was associated with medial forefoot pressures, especially during walking in habitually barefoot populations. Also, the midfoot shape metric (artificial intelligence) was associated with lateral forefoot pressures during walking in both habitually barefoot and shod populations. An improved predictive statistical model (support vector machine) also showed an increased prediction accuracy on the basis of the same data set (Xiang et al., 2022a). Rapid prediction of foot function from easily measured foot shape metrics may become the norm and extend into clinical diagnosis of foot pathologies and customized footwear development.

Conclusion and future perspectives

This review has explored known associations between foot shape, posture, and foot conditions. The foot artificial intelligence was found to be a useful metric of foot posture, which can be measured quickly and associated with motion and joint pain in adults; however, it may be less useful in special population groups, such as developing and obese individuals. The foot posture index to classify feet as pronated, neutral, and supinated is a useful approach under both static and dynamic conditions, but care must be taken when using it to associate pronation with injury risk. While recent studies have associated foot posture with joint loading, more information, such as factors of Achilles tendon angle and longitudinal arch angle, is needed to elucidate risk from normal function. With increasing imaging technology and database sharing, there is an opportunity to apply statistical shape modeling methods to derive learnt information from 'big data' and use this to make a population-based inference and fill in missing data from personalized information. Foot bone shapes and tissue morphology have been associated with pathology, gender, age, and height and may help develop rapid population-specific foot classifiers. A popular topic, barefoot running, was investigated, and it was shown that the forefoot toe shape influenced forefoot plantar pressure in habitually barefoot runners only, while the arch shape influenced plantar pressure in any population. This may play a role in footwear design.

Based on findings from the current review and potential gaps in the literature, future studies may consider the following topics *via* 1) tracking the internal foot motion during dynamic activities *via* biplanar fluoroscopy and multi-segmental models. With reported increased measuring accuracy, mapping the biplanar 2D motion to 3D shape motion and statistical shape modeling *in vivo* would assist the validation of finite element modeling and reveal the tissue variation, ligament strain, and cartilage loadings; 2) implementing different multivariate (support-vector) machine learning or convolutional neural network (CNN) algorithm to address potential nonlinear correlation scenarios in foot mechanics with shape or posture metrics; 3) standardizing data sets with synchronized IMU data for rapid prediction of instant mechanics, load accumulation, injury risks, and adaptation in the tissues and bones of the foot and correlating with foot shape; 4) analyzing the dynamic foot shape and posture *via* marker-less depth camera and real-time processing techniques under real-life scenarios for precise evaluation of clinical foot conditions and performance-fit footwear development.

Author contributions

Conceptualization: QM, HK, YG, and JF; methodology: QM, HK, and LX; visualization: QM, VS, and AW; writing—original draft preparation: QM, HK, and JF; writing—review and editing: VS, AW, and JB; supervision: YG and JF; project administration: QM, YG, and JF; funding acquisition: QM, JB, and YG. All authors have approved the submission of this manuscript.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Effects of visual deprivation on the injury of lower extremities among functional ankle instability patients during drop landing: A kinetics perspective

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Background: The ankle is prone to injury during drop landing with usual residual symptoms, and functional ankle instability (FAI) is the most common. Vision guarantees the postural stability of patients with FAI, and visual deprivation (VD) increases their risk of injury when completing various movements. This study explored injury risk during drop landing in patients with FAI under VD through the kinetics of lower extremities.

Methods: A total of 12 males with FAI participated in the study (age, 23.0 ± 0.8 years; height, 1.68 ± 0.06 m; weight, and 62.2 ± 10.4 kg) completed single-leg drop landings under visual presence (VP) and VD conditions. Ground reaction force (GRF), time to peak GRF, joint torque, and vertical length variation (ΔL) were measured.

Results: Significant effects were detected in the group for time to peak lateral GRF ($p = 0.004$), hip extensor torque ($p = 0.022$), ankle plantarflexion torque ($p < 0.001$), ankle varus torque ($p = 0.021$), lower extremity stiffness ($p = 0.035$), and ankle stiffness ($p < 0.001$). Significant effects of conditions were detected for vertical GRF, time to peak vertical and lateral GRF, loading rate, hip extensor torque, knee extensor torque, hip varus torque, knee varus torque, lower extremity stiffness, and ankle stiffness ($p < 0.05$). ΔL was affected by VD with a significant difference ($p < 0.001$).

Conclusion: In patients with FAI, an unstable extremity has a higher injury risk than a stable extremity, and VD increases such risk. However, because the influence of the central nervous system on hip strategy is also affected, the effect on the unstable extremity is more significant and more likely to result in injury. Deepening the squat range may be an effective preventive measure for reducing injury risk of unstable extremities during drop landing.

KEYWORDS

injury risk, motor strategies, lower extremities, biofeedback, drop landing

Introduction

Ankle sprain is the most common and severe injury in sports (Mckay et al., 2001). Repeated ankle injuries lead to chronic ankle instability (CAI), which is mainly caused by old injuries to ankle ligaments. CAI symptoms include local pain, decreased postural control, and decreased proprioception. CAI can be divided into functional ankle instability (FAI) and mechanical ankle instability (MAI) (Yu et al., 2022). Compared with individuals without a history of ankle injury, individuals with ankle injury history reported significant deficits in foot proprioception and in their static and dynamic balance (Alghadir et al., 2020), and over half of them did not seek professional treatment (Mckay et al., 2001). Muscle force and movement control are impaired if the injured ankle does not receive prompt and professional rehabilitation treatment, resulting in FAI (Freeman, 1965; Kim et al., 2021). Patients with FAI are almost five times more likely to sustain repeated episodes of ankle instability and recurrent ankle sprains than are healthy individuals (Mckay et al., 2001; Kaminski et al., 2003; Raymond et al., 2012; Gouttebarge et al., 2017), leading to irreversible damage to ankle stability (Hong et al., 2021).

Human sensory systems play essential roles in adjusting postural control and preventing injury, including the proprioception (Horak et al., 1990; Delahunt et al., 2006; Imbimbo et al., 2021; Oosterman et al., 2021), vestibular (Gosselin and Fagan, 2014; Brown et al., 2021; Kalinina et al., 2021; Morawietz and Muehlbauer, 2021), and visual systems. Patients with FAI usually exhibit proprioceptive impairment, which is the principal consideration for recurrent sprains, and visual information provides positional sensation and feedback to the motor system to offset deficits in proprioception and maintain their postural stability (Doherty et al., 2014; Toth et al., 2017; Raffalt et al., 2019; Han et al., 2022). Therefore, once patients with FAI transit from a visual presence (VP) to a visual deprivation (VD) condition, visual information cannot be transmitted and their postural stability becomes more challenging (Raffalt et al., 2019; Rosen et al., 2021). Some studies related to gait in VD suggest that VD is the main cause of falls in elderly individuals and of increased injury rates in athletes (Isakov and Mizrahi, 1997; Duggan et al., 2017; McDonough et al., 2017). Nonetheless, the research on VD among patients with FAI is relatively deficient, especially when patients are completing challenging tasks.

In addition, drop landing is one of the most challenging motion forms in daily physical activities and professional training. It is commonly seen in sports such as basketball and volleyball, and even in the process of stepping down during daily life. Ankle injuries from drop landing can be as high as 45%. Some studies have shown that patients with FAI have significantly decreased local muscle activity and deficits in postural instability (Delahunt et al., 2006; Rosen et al., 2013). Postural stability during drop landing requires rapid visual and position sensation integration. However, protective strategies in patients with FAI during drop landing with VD remain unclear, and the risk factors associated with VD in patients with FAI must be identified. This paper aimed to characterize the effects of VD and compensatory mechanisms under VD in patients with FAI during drop landing, through the kinetics of lower extremities. It hypothesized the following: 1) differences exist in the kinetics of patients with FAI during drop landing on stable and unstable extremities, 2) the injury risk among patients with FAI increases with VD during drop landing, and 3) VD affects unstable extremities during drop landing in patients with FAI.

Materials and methods

Participants

A total of 12 males with FAI participated in the study, and all provided informed consent. The Soochow University Ethics Committee Board approved this study. The demographic information of the participants is shown in Table 1. Self-reported instability and function were determined using the ankle joint functional assessment tool (AJFAT) (Ross et al., 2008; Ibrahim and Abdallah, 2020). Participants' inclusion criteria were as follows: 1) one unstable extremity, with AJFAT scores ≤26 (Delahunt et al., 2006; de Noronha et al., 2008; Kim et al., 2021; Rosen et al., 2021); 2) one stable extremity, with AJFAT scores >26; 3) no vestibular, visual, or neurological disease; 4) no lower extremity injuries for ≥6 months; and 5) only one side of the lower extremities suffering from FAI.

Experimental procedure

Participants had had no strenuous activities or muscle fatigue within 24 h before the experiment. Initially, participants warmed

TABLE 1 Sociodemographic characteristics at baseline.

Characteristic		Patients with FAI (n = 12)
Age (years)		23.00 ± 0.78
Height (m)		1.68 ± 0.06
Weight (kg)		62.15 ± 10.44
Unstable extremity	Left	9/75%
	Right	3/25%

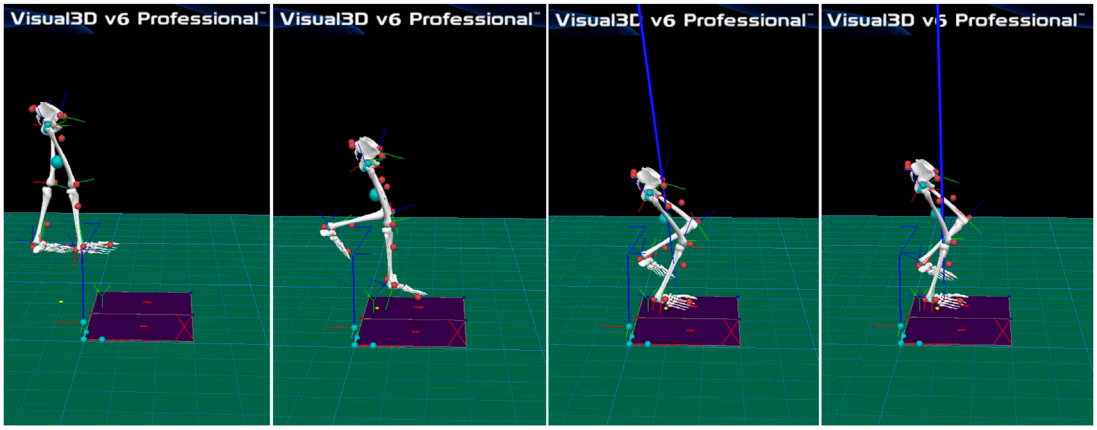

FIGURE 1
Participants landed toe first on one leg from a 30-cm-high step onto a force platform and were instructed to regain postural stability as quickly as possible.

up on a treadmill at 2.2 m/s for 3–5 min and stretched. In a single session, each participant with FAI wore experimental shorts and shoes and completed random drop landing trials without initial speed under four conditions, consisting of all combinations of each leg (unstable and stable extremities) and two conditions (VP and VD). Subsequently, participants completed a tiptoe slide from the platform's edge at a height of 30 cm, with instructions to place the hands akimbo to reduce the influence of arm swing. Two examiners provided safety assistance throughout the experiment. Landing on a single leg and bending knees instinctively and toe-first is illustrated in Figure 1. The two visual conditions were tested randomly. The participants were required to wear black blindfolds when testing under VD conditions (Shoja et al., 2020).

Each condition repetition was recorded by one examiner, who was blinded to the AJFAT scores, with a rest of 30 s between landings until the participant had completed three successful trials. Successful trials were defined as those performed without any extraneous movement during the 3 s prior to the hop and with the participant subsequently remaining upright during the 2 s after the drop landing, with the tested extremity within the boundary.

Data processing

Lower extremities' kinetic data were collected at 100 Hz using a 3D motion capture system equipped with eight infrared cameras (Vicon Motion Analysis, United Kingdom) by tracking 18 infrared-reflective balls (reflective markers) of 14 mm diameter. Ground reaction force data were sampled at 1000 Hz using a 3D Kistler force measuring platform (model: 9287B, 90 cm × 60 cm×10 cm, Kistler, Inc., Switzerland). The plug-in gait was modeled using Visual3D (Version 6, C-Motion, Inc., United States) to calculate joint kinetics.

The following kinetic data were analyzed: ground reaction force (GRF, including vertical GRF [*vGRF*], medial GRF [*mGRF*], and lateral GRF [*lGRF*]), time to peak GRF (T_GRF, including time to peak vertical GRF [*T_vGRF*], time to peak medial GRF [*T_mGRF*], and time to peak lateral GRF [*T_lGRF*]), loading rate (LR), joint torque in the sagittal plane and the frontal plane, hip joint vertical length variation (ΔL), lower extremity stiffness (K_{leg}), and ankle stiffness. LR was calculated using Eq. 1. K_{leg} was calculated using Eq. 2.

$$\mathbf{LR} = \mathbf{VGRF}/\mathbf{T_vGRF}, \quad (1)$$

$$\mathbf{K_{leg}} = \mathbf{vGRF}/\Delta\mathbf{L}. \quad (2)$$

Statistical analysis

The normality of each variable was determined using the Shapiro–Wilk test. Descriptive and dependent measures were calculated as mean ± SD. Dependent measures were used to examine the interaction effects of groups (unstable and stable extremities of patients with FAI) and conditions (VP and VD) that served as the repeated measure. Simple effect analysis was performed when significant interactions were found. If the simple effect analysis suggested a statistically significant difference in a metric in both groups between VP and VD, then the difference between VP and VD in each group was determined using the independent samples *t*-test. All statistical analyses were performed using a statistical software package (IBM SPSS Statistics, IBM, NY, United States), and the level of statistical significance was set at 0.05.

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TABLE 2 GRF, T-GRF, and LR of the unstable and stable extremities during landing with and without visual deprivation.

Variable	Group		Group	Condition	Condition × group
	Unstable extremity	Stable extremity	*p*-value	*p*-value	*p*-value
Direction of Ground Reaction Force (N)					
Vertical GRF					
VP	3.45 ± 0.48	3.62 ± 0.36	0.221	0.041	0.184
VD	3.92 ± 0.52	3.91 ± 0.46			
Medial GRF					
VP	0.27 ± 0.04	0.29 ± 0.02	0.311	0.959	0.647
VD	0.27 ± 0.03	0.28 ± 0.02			
Lateral GRF					
VP	−0.30 ± 0.02	−0.29 ± 0.02	0.999	0.195	0.999
VD	−0.30 ± 0.02	−0.29 ± 0.03			
Time to Peak Ground Reaction Force (ms)					
T_vGRF					
VP	59.00 ± 4.11	59.33 ± 2.93	0.229	0.001	0.186
VD	54.25 ± 5.60	57.17 ± 2.44			
T_mGRF					
VP	60.75 ± 3.52	60.92 ± 1.73	0.909	0.085	0.588
VD	60.08 ± 3.68	59.67 ± 2.46			
T_lGRF					
VP	58.17 ± 3.71	60.42 ± 1.83	0.004	<0.001	0.180
VD	54.33 ± 3.20	58.33 ± 2.31			
Loading rate (ms^{-1})					
VP	60.63 ± 8.56	61.12 ± 5.40	0.964	0.001	0.676
VD	66.69 ± 9.98	65.93 ± 7.18			

"–" represents the value of lGRF.

Results

GRF, T_GRF, and LR

The experimental data are shown in Table 2. No significant interaction effects in the group were detected for GRF, T_GRF, and LR. Significant effects of the condition were detected for vGRF ($p = 0.041$), T_vGRF ($p = 0.001$), T_lGRF ($p < 0.001$), and LR ($p = 0.001$). A significant effect in the group was detected for T_lGRF ($p = 0.004$).

Joint torque

The experimental data are shown in Table 3. No significant interaction effect between the group and the condition was detected for joint torque. Significant effects of the condition were detected for hip extensor torque ($p = 0.025$), knee extensor torque ($p = 0.016$), hip varus torque ($p = 0.025$), and knee varus torque ($p = 0.002$). Significant effects in the group were detected for hip extensor torque ($p = 0.022$), ankle plantarflexion torque ($p < 0.001$), and ankle varus torque ($p = 0.021$).

Stiffness

The experimental data are shown in Table 4. A significant interaction was detected for ΔL, and a statistically significant difference in ΔL was found between VP and VD in both groups. The ΔL of the groups decreased with VD, and VD affected unstable and stable extremities (unstable extremity, 0.236 ± 0.003 to 0.118 ± 0.003, $p < 0.001$; stable extremity, 0.256 ±

TABLE 3 Joint torque (hip, knee, and ankle) in the sagittal and frontal planes during landing with and without visual deprivation.

Variable	Group		Group	Condition	Condition × group
	Unstable extremity	Stable extremity	*p*-value	*p*-value	*p*-value
Joint torque in the sagittal plane (Nm/kg)					
Hip					
VP	−1.40 ± 0.23	−1.16 ± 0.40	0.022	0.025	0.461
VD	−1.64 ± 0.41	−1.29 ± 0.33			
Knee					
VP	1.32 ± 0.20	1.15 ± 0.29	0.090	0.016	0.913
VD	1.49 ± 0.29	1.34 ± 0.30			
Ankle					
VP	−1.76 ± 0.16	−1.54 ± 0.27	<0.001	0.825	0.091
VD	−1.90 ± 0.26	−1.44 ± 0.29			
Joint torque in the frontal plane (Nm/kg)					
Hip					
VP	−0.66 ± 0.12	−0.69 ± 0.14	0.398	0.025	0.205
VD	−0.88 ± 0.23	−0.76 ± 0.25			
Knee					
VP	0.05 ± 0.34	0.21 ± 0.50	0.223	0.002	0.757
VD	0.37 ± 0.51	0.59 ± 0.35			
Ankle					
VP	1.53 ± 0.20	1.40 ± 0.24	0.021	0.291	0.153
VD	1.81 ± 0.34	1.36 ± 0.62			

"–" represents hip extensor torque, ankle plantarflexion torque, hip varus torque, and knee eversion torque.

TABLE 4 Stiffness of FAI during landing with and without visual deprivation.

Variable	Group		Group	Condition	Condition × group
	Unstable extremity	Stable extremity	*p*-value	*p*-value	*p*-value
ΔL (m)					
VP	0.236 ± 0.003	0.256 ± 0.003	<0.001	<0.001	<0.001
VD	0.118 ± 0.003	0.127 ± 0.002			
K_{leg} *(BW/m)*					
VP	16.63 ± 2.19	14.56 ± 1.18	0.035	<0.001	0.693
VD	30.18 ± 3.20	28.52 ± 2.53			
*Ankle stiffness (N*m/Δθ)*					
VP	−0.038 ± 0.005	-0.032 ± 0.006	0.001	0.002	0.263
VD	−0.047 ± 0.011	-0.036 ± 0.004			

0.003 to 0.127 ± 0.002, $p < 0.001$). The unstable extremity revealed a shorter ΔL than the stable extremity, whether under VD or VP conditions (VP, 0.236 ± 0.003 to 0.256 ± 0.003, $p < 0.001$; VD, 0.118 ± 0.003 to 0.127 ± 0.002, $p < 0.001$). Furthermore, significant effects of the condition were detected for K_{leg} ($p < 0.001$) and ankle stiffness ($p = 0.002$). Significant effects in the group were detected for K_{leg} ($p = 0.035$) and ankle stiffness ($p < 0.001$).

Discussion

In this study, we focused on the effects of VD on patients with FAI during drop landing from the perspective of kinetics. We found differences between stable and unstable extremities in patients with FAI during drop landing. VD has a negative impact on patients with FAI, and the impact on the unstable extremity is more pronounced, which is consistent with our hypothesis.

Differences in kinetics between unstable and stable extremities

The results show that the T_lGRF was earlier in the unstable extremity than in the stable extremity, suggesting that the protective strategy for the unstable extremity may need to be sufficiently activated. Our results are similar to those of previous studies of patients with FAI during single-extremity jump landings, in that the severity of ankle instability correlates with the earlier frontal stabilization time (Ross et al., 2008). Some scholars (Lephart et al., 1998; Svinin et al., 2019) have suggested mechanisms responsible for the earlier T_lGRF in the unstable extremities and that the severity of ankle instability correlates with the shorter time to frontal stabilization. The proprioceptors in the ankle provide joint position sensation and motion signals along the sensory nerves to the spinal cord, where other sensory neurons may process and then transmit them to the brain for further processing. The proprioceptors damaged by recurrent sprains in the unstable extremities of patients with FAI may cause delayed signal transmission resulting in abnormal feedback (Lephart et al., 1998; Eechaute et al., 2009; Song et al., 2016; Ludvig et al., 2017; Ibrahim and Abdallah, 2020). This abnormal feedback may not be able to activate the protective strategy for landing in a timely, effective manner to reduce the injury risk. Moreover, studies (Horak et al., 1990; Patla et al., 2002; Chaudhry et al., 2005; Delahunt et al., 2006; Ergen and Ulkar, 2008; Kabbaligere et al., 2017; Ibrahim and Abdallah, 2020) have suggested that movement impairment could be modified by correcting muscle activations. Impaired proprioception causes a delay in the conduction of the afferent signal, and the consequent delay in the efferent signal causes a delay in the corrective muscle activation, which leads to a higher injury risk in the unstable extremity.

Higher hip extensor and ankle torque in the sagittal and frontal planes were revealed in the unstable extremity as compared with the stable extremity. Our results are similar to research on the effect of proprioceptive training on postural stability, that is, they show that a higher hip extensor torque increases ankle sprain risk (Hietamo et al., 2021). Joint torque represents the buffering function of energy absorption on the ground reaction force through muscle strength. The higher the hip extensor torque is, the less GRF is absorbed through the muscles, which means that the role of muscle strength in reducing ground reaction force is minor; the higher the load is, the less the extremity is buffered during drop landing. Based on kinetic chain theories, insufficient hip muscle function increases the likelihood of uncontrolled ankle displacements and ankle injury.

Previous studies have showed poor postural stability in ankle sprain patients (Delahunt et al., 2006; Doherty et al., 2014). Postural stability decreases when the unstable extremity is forced to support itself during drop landing. Corresponding to our results, studies have found that K_{leg} is larger in the unstable extremity than in the stable extremity. A study related to ankle and knee proprioception found that postural swing and center of body mass changes increase among ankle-injured individuals compared with the uninjured group and that a larger K_{leg} response might increase postural swing and center of body mass changes during movement (Lephart et al., 1998). Some studies (Ito and Gomi, 2020) have suggested the mechanisms of postural swing and center of body mass changes, in which impaired proprioception in the unstable extremity makes accurate adjustment signal transmission for joint position sense abnormal, then blocks joint pressure and tension feedback and further affects the nerve fiber processing of sensory information, thus aggravating postural instability.

The greater ankle stiffness in the stable extremity compared with the unstable extremity also supports the view that the unstable extremity has a higher injury risk. Ankle stiffness reduces energy consumption caused by joint activity through the rigid lever, and stored energy is released to cushion the ground impact force; ankle stiffness can prevent joint instability during drop landing (Gao et al., 2022). Ludvig et al. (2017) has suggested mechanisms for deficient levels of ankle stiffness, in which proprioception is impaired in the unstable extremity, leading to the muscle not activating. The ligaments and tendons become slack due to recurrent ankle sprain, resulting in energy consumption caused by joint activity. The energy available for cushioning the ground impact force is reduced, causing higher injury risk.

Effect of visual conditions on patients with FAI

The result of increased vGRF and LR under VD in our paper shows the higher impact force and injury risk in the lower extremities of patients with FAI during drop landing with VD

(Rosen et al., 2013). Under VD, the protective strategy may not be sufficiently activated to buffer load and reduce injury risk. Our results show that earlier T_vGRF and T_lGRF appear under VP than under VD. Previous studies have found that visual conditions affect postural stability (Zipori et al., 2018), and our research further found that VD increases the risk of injury in patients with FAI. Under VD, the brain is unable to conduct environmental image analysis and deliver control signals generated by visual information transfer to achieve a series of postural control outputs, such as muscle tension adjustment and joint angle change.

Moreover, our results show deficits in postural control in patients with FAI, revealing that hip and knee torque in the sagittal and frontal planes increase with VD and that the hip and knee strategies mobilize when the ankle is unresponsive. This phenomenon shows that the dynamic defense system of the joint is further limited due to the proprioception impairment caused by the recurrent ankle sprain, which leads to abnormal ankle strategies in postural control (Alghadir et al., 2020). In our paper, neither the unstable nor the stable extremities buffered the load through ankle torque. This result could reflect the inability of the unstable extremity to overcome the loss of two sources of sensory information (Song et al., 2016): proprioception, which is altered due to repeated ankle sprain, and visual inputs, which are lost under VD. Palm et al. (2009) studied the effects of vision and hearing on postural stability and suggested that other senses, such as proprioception and vestibular sense, can compensate for postural instability when visual information input is abnormal. Zipori et al. (2018) also suggested that proprioception dependence in individuals increases when visual information inputs are abnormal. However, our paper found that even the stable extremity, without proprioception impairment, does not show special performance compared with the unstable extremity during VD. Therefore, as for the stable extremity, proprioception dependence is reduced under the long-term influence of the unstable extremity deficits (Eechaute et al., 2009; Song et al., 2016).

Our results show that K_{leg} is larger under VD than under VP, indicating an increased instability and injury risk in the lower extremities. As mentioned previously, a larger K_{leg} may increase center of body mass changes and postural swing during drop landing. Eechaute et al. (2009) have supported this view and suggested that the center of body mass is used to maintain balance during motion. Palm et al. (2009) suggested that when visual control information decreases, postural control ability decreases and the lower extremity injury risk increases accordingly. In addition, muscle stiffness regulation is the first reflex mechanism to adjust posture stability. Neurofeedback modulates systems and adjusts muscle tone to adapt to changes in the spatial environment. A possible explanation for the lower ankle stiffness under VD could be that missed visual inputs impede information transmission, which activates the muscles (McKeon and Hertel, 2008; Ibrahim and Abdallah, 2020). Miller et al. (2018) have further reported the importance of visual feedback, which may allow users to receive external joint position cueing, which is also conducive to sending the protection strategy's activation signal. The motion system subsequently manipulates ankle positioning and stiffness. When patients with FAI lose their normal perception of the external environment, ankle stiffness cannot be adjusted correctly.

Practical implications

Patients with FAI are highly dependent on vision in postural control. Vision plays an essential role in protective strategy activation for providing assistance information during drop landing among patients with FAI. The results reported here show that the ΔL adopted is larger under VP than under VD. The ΔL is the vertical length change of the hip, reflecting the cushioning ability of lower extremities to the ground impact force, which could be considered a hip strategy that patients with FAI enable. The visual system supplies the central nervous system with continuous information about the body's position in the environment and activates the protection strategy. The hip strategy undertakes most of the buffering load tasks. Studies of postural strategies related to the loss of proprioception and vestibular function have found that the hip strategy is adopted more when joint instability occurs (Horak et al., 1990; Lee and Bo, 2021; Lee et al., 2022). Furthermore, the body mainly relies on the ankle strategy to adjust postural stability when interferences are few or when environmental information changes, but the hip strategy also dominates for large changes. Therefore, the human body prioritizes adjusting hip activities to reduce the load during drop landing, such as by increasing hip flexion to maintain postural stability. One study (Palm et al., 2009) has indicated, by analyzing biofeedback mechanisms through a trial of the effects of visual and auditory inputs on posture control, that postural stability increases with an increasing degree of visual control. The unstable extremity relies more on visual inputs for postural control than does the stable extremity. In support, studies related to visual use have suggested an explanation for postural control deficits in patients with FAI: their reliance on visual information has increased due to decreased somatosensory information from the ankle complex (Song et al., 2016; Nobusako et al., 2021). Our results are similar, and we found a shorter ΔL on the unstable and stable extremities under VD than under VP. Still, the unstable extremity was shorter, indicating that vision plays an important role in maintaining the postural stability of patients with FAI; the effect of VD on the unstable extremity is more significant, which means that the hip strategy on the unstable extremity is more difficult to activate through increasing hip and knee flexion. Furthermore, in our study, lower ankle stiffness, in which the ankle cannot form a more stable structure, was revealed under VD. However, a more vertical and protective landing strategy to minimize injury risk has been proposed that provides greater

mechanical protection by generating a close-packed position in the ankle or by increasing hip flexion to provide stability for an uncertain landing. To sum up, increasing hip flexion to increase ΔL by deepening squat range might be an effective precaution in reducing injury risk in the unstable extremity during drop landing. Miller et al. (2018) have found that individuals with lower ankle stiffness would utilize a more protective landing strategy that increases hip and knee flexion to provide stability for an uncertain landing.

In addition, Ibrahim and Abdallah (2020) have suggested that proprioception training improves balance and shortens reaction time in postural stability, which further proves that the later T_lGRF in the stable extremity in our paper arises from self-protection. Hence, the extension of T_GRF can also be used as a breakthrough in precautions for patients with FAI. The current paper also mentions the benefits of repeated practice. As one repeats these activities, a feedforward mechanism occurs unconsciously through the activation of the stabilizing joint muscles and through the buffered load by activated muscle strategies absorbing energy.

Limitations

As an indicator of postural stability, ankle muscle strength plays a vital role in postural stability. This limitation is also a characteristic of our paper, which evaluated only kinetics during this protocol, so the possibility of an unknown kinetic effect of the muscle activation response on the lower extremities cannot be determined. Therefore, surface electromyography (sEMG) could be combined in future research on patients with FAI under VD to validate the dependence on vision and to determine what strategies the muscles take, whether protection strategies through the muscles are adopted by the unstable extremity to reduce the risk of injury, and whether warming up in advance can help muscles to activate faster as a precaution against ankle injury.

Conclusion

Unstable extremities have additional injury risks compared with stable extremities in patients with FAI, and vision plays an important role in maintaining postural stability in these patients. VD increases the injury risk for both the unstable and stable extremities, but the effect on the unstable extremities is more significant and more likely to result in injury, as the control of the central nervous system on hip strategy is also affected. Deepening the squat range and avoiding prolonged reaction time might be effective precautions for reducing injury risk in patients with FAI during drop landing. In the future, muscle strength and training could be studied and combined with sEMG to evaluate precautions from a muscular perspective.

Data availability statement

The original contributions presented in the study are included in the article/Supplementary Material; further inquiries can be directed to the corresponding authors.

Ethics statement

The studies involving human participants were reviewed and approved by the Soochow University Ethics Committee Board. The patients/participants provided their written informed consent to participate in this study.

Author contributions

LM, LK, and LK conceived the presented idea, developed the framework, and wrote the manuscript. QZ, JS, and YH provided critical feedback and contributed to the final version. All authors were involved in the final direction of the paper.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Effects of traditional Chinese herb hot compress combined with therapeutic exercise on pain, proprioception, and functional performance among older adults with knee osteoarthritis: A randomized controlled trial

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Background: Knee osteoarthritis (KOA) is one of the most common chronic progressive diseases with degenerative destruction of articular cartilage and bone, leading to knee pain, impaired proprioception, and reduced functional performance. This study was to investigate the effects of an 8-week Traditional Chinese herb hot compress (TCHHC) combined with therapeutic exercise (TE) on pain, proprioception, and functional performance among older adults with KOA.

Methods: Twenty-seven older adults with KOA were recruited and randomly assigned to the TCHHC + TE or TE groups. Thirteen participants received TCHHC + TE, and fourteen received TE. At pre- (week 0) and post-intervention (week 9), their pain, joint proprioception, and functional performance were measured. Two-way ANOVA with repeated measures was adopted to analyze the data.

Results: Compared with week 0, the pain score, proprioception thresholds of knee extension and ankle plantarflexion, and the times of TUG and 20-m walk tests decreased more significantly in the TCHHC + TE group than in the TE group at week 9.

Conclusion: Compared with TE, the 8-week TCHHC + TE was superior in relieving pain, recovering proprioception, and improving functional performance among older adults with KOA. It is recommended that TCHHC should be adopted prior to TE to enhance the effects of KOA rehabilitation.

KEYWORDS

osteoarthritis, pain, proprioception, functional performance, Chinese herbs, therapeutic exercise

1 Introduction

Knee osteoarthritis (KOA) is one of the most common chronic progressive diseases, with a prevalence of 30%–40% among older adults over 65 years of age, posing a serious health risk (Helmick et al., 2008). KOA is associated with degenerative destruction of articular cartilage and bone, leading to knee pain (Glyn-Jones et al., 2015), impaired proprioception (Segal et al., 2010), and reduced functional performance (Hawker, 2019).

Pain is the predominant and most common clinical symptom of KOA, which was believed to be caused by the uneven load at the knee and the increased pressure on the tibiofemoral joint (Beaupré et al., 2000). Pain has been commonly assessed by the Western Ontario and McMasters Osteoarthritis Index (WOMAC) (Sisante et al., 2020). Proprioception is closely related to the developmental process of KOA (van der Esch et al., 2007), and the impairment of proprioception of the knee and ankle joints led to decreased lower extremity joint balance and coordination among older adults with KOA (Hassan et al., 2001; Knoop et al., 2011). Although KOA affects mostly the knee joint, ankle proprioception is closely related to balance control (Han et al., 2015). Compared with their healthy counterparts, older adults with KOA have worse functional performance, which affects their ability of daily activities and increases the risk of falls (Anderson et al., 2019). Timed up & Go (TUG) (Podsiadlo and Richardson, 1991; Dobson et al., 2012) and 20-m walk tests (Motyl et al., 2013) have been used effectively to quantify functional performance among older adults with KOA.

Older adults with KOA typically receive surgical, pharmacological, and non-pharmacological treatments (Brophy and Fillingham, 2022). Surgery is generally used in the late-stage and can improve symptoms of KOA (Xue et al., 2022), however, with a high cost (Serikova-Esengeldina et al., 2022); Non-steroidal antiinflammatory, opioids, or nutritional drugs are usually used in pharmacological treatments (Persson et al., 2018), and side effects on the liver and kidneys have been reported (Zhang et al., 2019b). Non-pharmacological treatments have been recommended as the primary treatments for KOA (Brophy and Fillingham, 2022). Among them, therapeutic exercise (TE), such as resistance training or joint mobilization, has the potential effects to relieve pain (Juhl et al., 2014), recover proprioception (Lai et al., 2018), and improve functional performance (Fransen et al., 2002). Resistance training improves muscle strength and functional performance (de Rooij et al., 2017). Joint mobilization balances the load at the knee and relieves pain (Chen H. et al., 2019), as well as improves proprioception by enhancing sensory input to the central nervous system about postural movements (Saunders et al., 2005). However, a previous meta-analysis study showed that the effect size of TE remained small to moderate (Fransen et al., 2015). Therefore, there is room to improve its effectiveness on KOA.

Traditional Chinese herb hot compress (TCHHC) has been widely used in the treatment of chronic diseases such as KOA (Yuan et al., 2015). It fully synergizes the warming effect, which has been shown to slightly improve pain, functional performance and quality of life in older adults with KOA (Aciksoz et al., 2017), and the pharmacological effect, which reduces inflammation for pain relief by promoting synovial fluid flow and reducing the release of inflammatory substances into the synovial fluid (Wang et al., 2010). Combined with the above-mentioned effects of resistance training and joint mobilization, TCHHC + TE has the potential to relieve pain, improve proprioception, and functional performance.

To the best of our knowledge, no studies have explored the effects of TCHHC + TE on pain relief, proprioception recovery, and functional performance improvement among older adults with KOA. It was hypothesized that compared with TE, TCHHC + TE was superior in relieving pain, recovering proprioception, and reducing the times of TUG and 20-m walk tests.

2 Material and methods

2.1 Sample size estimates

The sample size was estimated by An *a priori* power analysis (G*Power Version 3.1). Based on a previous report compared the pain score (interaction <0.001, $\eta 2_p = 0.638$) and proprioception (interaction = 0.006, $\eta 2_p = 0.267$) among older adults with KOA before and after a TE therapy or health lecture series (Shen et al., 2021). By setting the significance level to 0.05 and the statistical power to 80%, the minimum total sample size of this study should be 6 (calculated by pain score) and 26 (calculated by proprioception), respectively.

2.2 Participants

All participants were recruited and all the data were collected in the Neck, Shoulder, Back, and Leg Pain Hospital, Jinan, China, from June 2020 to November 2021. The inclusion criteria included the following: 1) 65 years old or older; 2) at least one knee was diagnosed with KOA according to the clinical criteria of the American College of Rheumatology; 3) a radiographic grade of 2 or higher by the Kellgren/Lawrence scale (K/L). The exclusion criteria included the following: 1) had a neurodegenerative disease or neurosensory disorder affecting the knee other than the KOA; 2) had a traumatic injury of the lower extremity joint in the past 3 months; 3) had planned for a total knee replacement in the following months; 4) had a history of allergies to Traditional Chinese herbs; 5) had received other KOA treatments, including steroids, intra-articular injection or other analgesic drugs within previous 3 months.

Thirty-four participants who met the above criteria were assigned to the TCHHC + TE or the TE groups at a ratio of 1:1 by computer-generated randomization procedure. Allocation

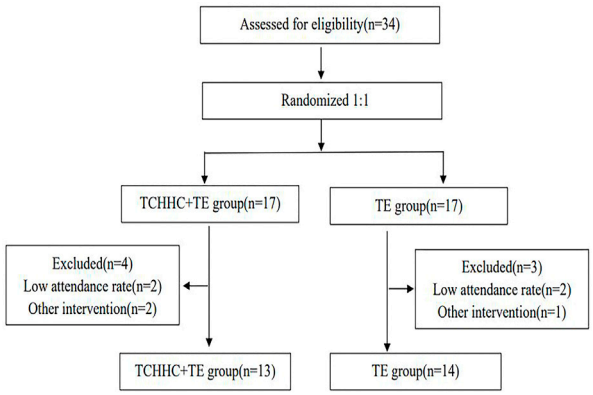

FIGURE 1
Chart flow.

concealment was ensured because allocation information was protected in opaque sealed envelopes and kept by investigators involved in participant recruitment. This study was single-blinded, with the investigators were blinded and participants knew the randomized assignments.

The participants in the TCHHC + TE group received TCHHC + TE, and those in the TE group received TE for 8 weeks. The participants were excluded with below 80% attendance rate (Song et al., 2020). The attendance rate is calculated by dividing the number of sessions completed by each participant by the total number of sessions. In the TCHHC + TE group, two participants were excluded due to their lower attendance rate (73% and 70%), and two were excluded due to co-intervention. In the TE group, two were excluded due to their lower attendance rate (74% and 72%), and one was excluded due to co-intervention. The reasons for failing to attend were transportation difficulties, bad weather, and family commitments. Co-intervention including steroids, intra-articular injections or other pain medications. Final analyses were conducted among thirteen participants in the TCHHC + TE group and fourteen in the TE group (Figure 1).

All participants were required to sign informed consent before participation. The project was approved by the Ethics Committee of Neck, Shoulder, Back, and Leg Pain Hospital (2021009) and conformed to the guidelines of the Declaration of Helsinki, and was registered in the Chinese Clinical Trial Registry with a registration number of ChiCTR2100052450.

2.3 Interventions

Both groups received three sessions (Tuesday, Thursday, and Saturday) of the intervention per week for 8 weeks. Participants were tested successively on Monday of the ninth week. In each session, the participants in the TCHHC + TE group received 30 min of TCHHC, 30 min of joint mobilization, and 30 min of resistance training. The TE group received 30 min of joint mobilization, 30 min of resistance training, and 30 min of health lecture series.

The dosage of the herbs was confirmed by the Apoplectic Joint Disease Pulse Syndrome and Treatment, Guizhi Shaoyao Zhimu Decoction, prescribed by Zhang Zhongjing, a famous Chinese medical practitioner of the Eastern Han Dynasty (from about 150 to 154 to about 215 to 219, A.D.) (Daily et al., 2017) and provided by the Neck, Shoulder, Back, and Leg Pain Hospital. The types and dosages of hot compress herbs are shown in Table1. The herbs were ground into 0.2 mm power in the hospital by the Dade medicine machine (model DF-20, Wenling Linda Machinery Co., Ltd, China), and mixed in a cloth bag of about 15 cm length*15 cm width after heating in a microwave oven for 40 s to make the temperature of the medicine bag reach 40–42°C, measured using the Raytek MiniTemp™ non-contact infrared thermometer (Raytek, United States). After cleaning the skin with saline gauze, the bag was quickly applied to the knee joint for about 30 min (Figure 2A).

The TE intervention included two parts, namely, resistance training (static and dynamic) and joint mobilization. During static resistance training, the participants lay in a supine position, contracted their quadriceps to keep their heels slightly off the bed for 10 s, and relaxed for 10 s, repeating about 50 reps per session. During dynamic resistance training, the participant sat in a training chair and flexed and extended the knee at approximately 90° to 170° with a weight adjustable bag (between 9.8 and 15N depending on the participant's knee recovery) strapped to the ankle to increase knee extension resistance for about 100 reps per session (Figure 2B). The joint mobilization was based on Maitland's method in Australia (Courtney et al., 2010). The participants lay in a supine position and received joint mobilization conducted by physical therapists about 8 times per session. The joint mobilization technique involved the following: medial and lateral sliding of the patella, and up and down, anterior and posterior sliding of the tibiofibular joint, anterior and posterior sliding of the tibiofemoral joint, and long-axis traction (Figure 2C).

All participants in the TE group participated in the health lecture series for 8 weeks, three sessions per week. One session of health lecture lasted for 30 min, including knowledge about KOA, awareness of chronic diseases, psychological health education, nutritional meals, scientific exercise, and exchange of experience. The format of the lecture was to watch selected TV programs or read related magazines.

2.4 Testing protocol

2.4.1 Pain

The pain score of the more affected lower extremity was assessed by five pain items of the WOMAC before and after the 8-week intervention, and its validity and reliability have been demonstrated (Bellamy et al., 1988). In each item, 0 points

TABLE 1 Components of Traditional Chinese herbs.

Components	Amount (g)	Components	Amount (g)
Cassia twig	30	White mustard seed	30
Paeony root	20	Sichuan ox knee	25
Myrrh	20	Mulberry parasitic	25
Ephedra root	20	Parsnip	20
Atractylodes macrocephala	20	Hot dog ridge	30
Frankincense	20	Antler gum	30
Prepared aconite	25	Anemarrhena asphodeloides	20
Prepared Sichuan black	25		

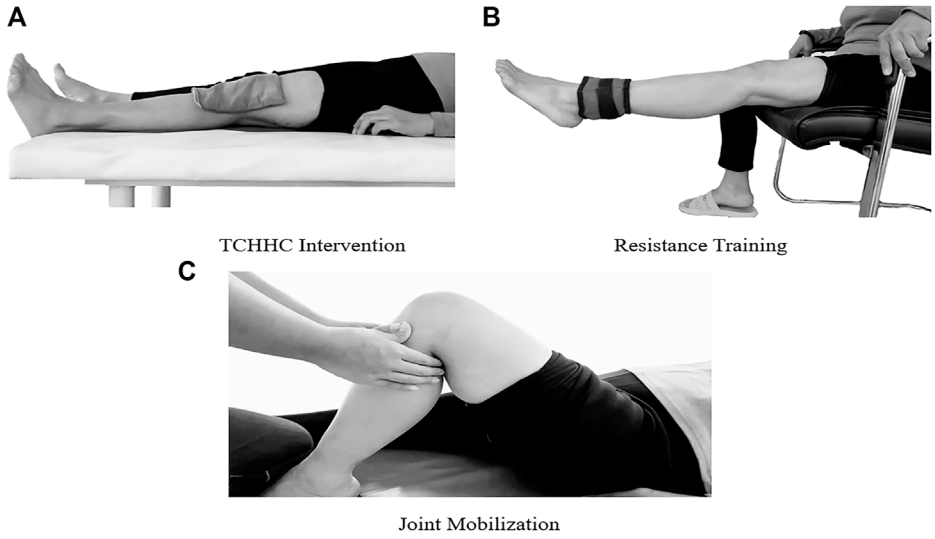

FIGURE 2
Illustrations of TCHHC **(A)**, resistance training **(B)**, and joint mobilization **(C)**.

represented "no pain," whereas 10 points represented "the worst pain possible". Higher scores indicate more severe pain.

2.4.2 Proprioceptive

The proprioceptive thresholds of knee flexion/extension, ankle dorsi/plantarflexion, and inversion/eversion of the more affected lower extremity were measured by proprioception devices (Sunny, AP-II, China). The test showed a good two-time-point intraclass correlation coefficient ((ICC = 0.737–0.935) (Sun et al., 2015). The proprioception devices collected the minimum angular motion that the patient was able to detect during knee flexion/extension, ankle dorsi/plantarflexion, and inversion/eversion. The devices consist of a platform that can rotate within the frontal and sagittal planes. The platform is driven by 2 electric motors at an angular velocity of 0.4°/s. An electronic goniometer in the devices recorded the angular displacement of the platform. Each participant was seated on a height-adjustable chair with their foot placed on the platform. During the ankle proprioception test, the knee and hip joints were flexed at 90°, and the leg was perpendicular to the surface of the platform when the platform was placed in a horizontal position. During the knee proprioception test, the lateral axis of the instrumentation was parallel with the mediolateral axis of the knee joint. The hip and knee joints were each positioned at 90°, whereas the ankle joint was at the neutral position. Approximately 50% subject's lower extremity weight was rested on the platform by using the thigh cuff suspension system to control unwanted sensory cues from the contact between the platform and the plantar surface of the foot. The participant sat with their eyes closed and wore headphones with music playing to eliminate potential environmental visual and auditory stimulation. The participant was instructed to concentrate on their foot and to

press the hand switch to stop the movement of the platform when they could sense motion followed by identification of the rotation direction. The motor was operated to rotate with a random time interval ranging from 2 s to 10 s after an indication to start a trial. At least five trials were performed for each direction to reduce random measurement errors (Song et al., 2021).

2.4.3 TUG test

The TUG test showed good reliability between raters (ICC = 0.99) and within the same raters on two consecutive tests (ICC = 0.99) (Podsiadlo and Richardson, 1991). The participants were seated in a 46 cm high seat. When the assistant gave the "go" command, the participants stood up and moved forward a 3-m distance as fast as possible, then turned around and sat back down. The time was taken from when they left the chair until they returned to the chair, in seconds. Shorter time represents better functional performance. Three successful trials were collected.

2.4.4 20-m walk test

The 20-m walk test showed good two-time-point reliability (Spearman r = 0.94–0.99) (Motyl et al., 2013). The participants walked as fast as possible within a 20-m distance. Shorter time represents better functional performance. Three successful trials were collected.

2.5 Statistical analysis

All variables were statistically analyzed using SPSS 26.0 (IBM SPSS, Armonk, NY, United States). The normality of all outcome variables was tested using Shapiro-Wilk tests. Two-way (group by time) ANOVAs with repeated measures were used to test the differences in dependent variables before and after the intervention. If significant interactions were detected, the stratified t-tests with the Bonferroni adjustment were performed. If there is no interaction, pre-and post-tests were compared by combining the two groups. Partial eta squared ($\eta 2_p$) was used to represent the effect size of the two-way ANOVA's main effects and interactions. The thresholds for $\eta 2_p$ were as follows: 0.01–0.06 for small, 0.06–0.14 for moderate, and>0.14 for large (Ledolter and Kardon, 2020). Cohen's *d* was used to represent the effect size of *post hoc* pairwise comparisons (Cohen, 1988). The thresholds for Cohen's *d* were as follows: <0.20 for trivial, 0.21–0.50 for small, 0.51–0.80 for medium, and >0.81 for large (J, 1988). The significance level was set at 0.05 (Ledolter and Kardon, 2020).

3 Results

No side effects were observed during the intervention, e.g., pain exacerbation, allergic reaction, redness, or heat burn due to the hot compress on the superficial joints. The Shapiro-Wilk test confirmed that all dependent variables were normally distributed. Independent t-tests showed no significant differences in age ($p = 0.794$), weight ($p = 0.962$), height ($p = 0.467$), body mass index ($p = 0.841$), and leg length ($p = 0.639$) between the two groups (Table 2).

The pain score was presented in Table 3. Significant interactions were detected in 2 items (walk on the ground: $p = 0.009$, $\eta 2_p = 0.245$ and go up or down stairs: $p = 0.006$, $\eta 2_p = 0.266$) of the WOMAC scores, and the total score ($p = 0.047$, $\eta 2_p = 0.148$). At week 9, the go up or down stair ($p = 0.002$, $d = 1.327$) and total ($p = 0.035$, $d = 0.798$) scores were lower in the TCHHC + TE group compared with those in the TE group. Compared to week 0, the sleep at night ($p < 0.001$, $\eta 2_p = 0.395$), sit or lying ($p = 0.001$, $\eta 2_p = 0.364$), and stand straight ($p < 0.001$, $\eta 2_p = 0.424$) scores were lower at week 9 in both groups.

The proprioceptive thresholds were presented in Table 4. Significant interactions were detected in proprioception thresholds of knee extension ($p = 0.018$, $\eta 2_p = 0.203$) and ankle plantarflexion ($p = 0.010$, $\eta 2_p = 0.235$). Post hoc comparisons showed that compared with week 0, the proprioception thresholds of the knee extension and ankle plantarflexion of both groups (TCHHC + TE: $p < 0.001$, $d = 1.870$, $p < 0.001$, $d = 2.434$; TE: $p = 0.007$, $d = 0.927$; $p < 0.001$, $d = 1.352$) decrease at week 9. The proprioception threshold of the knee extension was lower in the TCHHC + TE group compared with those in the TE group at week 9 ($p = 0.022$, $d = 0.967$). Significant time effects were detected in proprioception thresholds of knee flexion ($p < 0.001$ $\eta 2_p = 0.540$), ankle dorsiflexion ($p < 0.001$ $\eta 2_p = 0.446$), ankle inversion ($p = 0.048$, $\eta 2_p = 0.148$) and eversion ($p = 0.008$, $\eta 2_p = 0.246$).

The times of TUG and 20-m walk tests were presented in Table 5. Significant interactions were detected in the TUG ($p = 0.046$, $\eta 2_p = 0.150$) and 20-m walk ($p = 0.008$, $\eta 2_p = 0.249$) tests. Post hoc comparisons showed that compared with week 0, the times of the TUG and 20-m walk tests of both groups (TCHHC + TE: $p < 0.001$, $d = 4.420$, $p < 0.001$, $d = 3.400$; TE: $p < 0.001$, $d = 3.113$; $p < 0.001$, $d = 2.543$) decrease at week 9. The times of the TUG and 20-m walk tests was lower in the TCHHC + TE group compared with those in the TE group at week 9 ($p = 0.022$, $d = 0.957$; $p = 0.044$, $d = 0.820$).

4 Discussion

This study investigated the effects of an 8-week TCHHC + TE on pain, proprioception, and functional performance among older adults with KOA. The results supported our hypothesis. Compared with TE, TCHHC + TE was superior in relieving pain, recovering proprioception, and improving functional performance among older adults with KOA.

At week 9, pain scores decreased in both groups, and the decrease was more significant in the TCHHC + TE group than in the TE group. The effects of TE on pain relief have been well documented (Juhl et al., 2014; Zeng et al., 2021). Resistance

TABLE 2 Demographic characteristics of practitioners.

	TCHHC + TE (n = 13)	TE (n = 14)	p-value
More affected leg	5R, 8L	6R, 8L	--
K/L	4 II, 7 III, 2 IV	4 II, 8 III, 2 IV	--
Age (y)	72.0 ± 5.4	72.6 ± 5.9	0.794
Weight (kg)	69.9 ± 5.2	70.0 ± 7.7	0.962
Height (cm)	162.2 ± 4.1	163.6 ± 5.7	0.467
BMI (kg/m^2)	26.7 ± 2.2	26.5 ± 2.5	0.841
Leg length (cm)	82.3 ± 3.1	81.7 ± 3.4	0.639

TABLE 3 Pain score of the more affected at weeks 0 and 9.

Pain score		TCHHC + TE (n = 13)	TE (n = 14)	Time		Group		Time[a] group		Post hoc	
				p	η_p^2	p	η_p^2	p	η_p^2	p	d
Walk on the ground	Week 0	5.08 ± 2.25	4.95 ± 1.38	--	--	--	--	0.009	0.245	[a1] < 0.001,[a2] < 0.001	[a1] = 1.605,[a2] = 1.874
	Week 9	1.85 ± 1.63[a1]	2.50 ± 1.22[a2]							--	--
Go up or down stairs	Week 0	6.62 ± 2.22	6.61 ± 1.75	--	--	--	--	0.006	0.266	[a1] < 0.001,[a2] < 0.001	[a1] = 2.316,[a2] = 1.822
	Week 9	2.15 ± 1.28[a1b]	3.79 ± 1.19[a2]							[b] = 0.002	[b] = 1.327
Sleep at night	Week 0	4.23 ± 2.65	4.36 ± 2.27	<0.001	0.395	0.596	0.011	0.302	0.043	--	--
	Week 9	1.31 ± 1.25	1.64 ± 1.40							--	--
Sit or lying	Week 0	2.85 ± 1.95	2.64 ± 1.28	0.001	0.364	0.835	0.002	0.745	0.004	--	--
	Week 9	1.23 ± 0.93	1.29 ± 1.14							--	--
Stand straight	Week 0	2.62 ± 1.80	2.92 ± 1.77	<0.001	0.424	0.059	0.135	0.257	0.051	--	--
	Week 9	1.38 ± 0.96	1.79 ± 1.37							--	--
Total score	Week 0	22.38 ± 7.43	20.35 ± 6.07	--	--	--	--	0.047	0.148	[a1] < 0.001,[a2] < 0.001	[a1] = 2.246,[a2] = 1.760
	Week 9	7.92 ± 3.92[a1b]	11.00 ± 3.80[a2]							[b] = 0.035	[b] = 0.798

[a]Denotes significant different between weeks 0 and 9 in each group.
[b]Denotes significant difference between the two groups at week 9.
TCHHC: traditional chinese herb hot compress.
TE: therapeutic exercise.

training increases the muscle strength of the lower extremity, improved the stability of knee joints, and reduces the wear between articular cartilage to relieve pain (Chen H. et al., 2019). Joint mobilization reduces the excitability of nerves to relieve pain (Saunders et al., 2005). Our results showed that TCHHC + TE significantly relieved pain among older adults with KOA compared to TE. Similar results were reported, supporting that the addition of TCHHC to TE enhanced the relief of knee pain (Chen S. et al., 2019). There are several possible mechanisms for TCHHC + TE's superior effects for pain relief, as follows. 1. Cao (Cao et al., 2016) et al. reported that the herbs have been demonstrated to be effective at promoting absorption of inflammatory substances and eliminating swelling by increasing blood circulation in clinical observation. In this study, some of the components of the herbs are effective in relieving pain, e.g., peony root reduces the expression level of substance P, which is the most important element in pain perception (Wang et al., 2013). Prepared sichuan black and

TABLE 4 Proprioception thresholds of the more affected knee joint, ankle sagittal plane, and ankle frontal plane at weeks 0 and 9.

Proprioception threshold (°)		TCHHC + TE (n = 13)	TE (n = 14)	Time		Group		Time[a] group		Post hoc	
				p	η_p^2	p	η_p^2	p	η_p^2	p	d
Knee flexion	Week 0	3.29 ± 1.57	2.34 ± 1.04	<0.001	0.540	0.095	0.108	0.296	0.043	--	--
	Week 9	1.76 ± 0.51	1.46 ± 0.69							--	--
Knee extension	Week 0	4.29 ± 1.72	3.31 ± 1.64	--	--	--	--	0.018	0.203	^{a1} < 0.001,^{a2} = 0.007	^{a1} = 1.870,^{a2} = 0.927
	Week 9	1.32 ± 0.31^{a1b}	1.99 ± 0.93^{a2}							^{b} = 0.022	^{b} = 0.967
Ankle plantarflexion	Week 0	4.40 ± 1.29	3.53 ± 1.22	--	--	--	--	0.010	0.235	^{a1} < 0.001,^{a2} < 0.001	^{a1} = 2.434,^{a2} = 1.352
	Week 9	1.68 ± 0.61^{a1}	2.10 ± 0.65^{a2}							--	--
Ankle dorsiflexion	Week 0	3.10 ± 1.13	3.22 ± 1.26	<0.001	0.446	0.479	0.020	0.674	0.007	--	--
	Week 9	1.95 ± 0.70	2.27 ± 0.80							--	--
Ankle inversion	Week 0	5.72 ± 2.44	6.04 ± 3.65	0.048	0.148	0.670	0.007	0.921	0.000	--	--
	Week 9	4.54 ± 2.50	4.97 ± 1.67							--	--
Ankle eversion	Week 0	5.51 ± 2.58	5.71 ± 2.93	0.008	0.246	0.268	0.049	0.198	0.065	--	--
	Week 9	3.51 ± 1.21	4.98 ± 1.99							--	--

[a]Denotes significant different between weeks 0 and 9 in each group.
[b]Denotes significant difference between the two groups at week 9.
TCHHC: traditional chinese herb hot compress.
TE: therapeutic exercise.

TABLE 5 The times of TUG and 20-m walk tests at weeks 0 and 9.

Variables		TCHHC + TE (n = 13)	TE (n = 14)	Time		Group		Time[a] group		Post hoc	
				p	η_p^2	p	η_p^2	p	η_p^2	p	d
TUG (s)	Week 0	11.67 ± 1.07	11.48 ± 1.14	--	--	--	--	0.046	0.150	^{a1} < 0.001,^{a2} < 0.001	^{a1} = 4.420,^{a2} = 3.113
	Week 9	7.41 ± 0.80^{a1b}	8.23 ± 0.91^{a2b}							^{b} = 0.022	^{b} = 0.957
20-m walk (s)	Week 0	16.61 ± 1.95	15.73 ± 1.58	--	--	--	--	0.008	0.249	^{a1} < 0.001,^{a2} < 0.001	^{a1} = 3.400,^{a2} = 2.543
	Week 9	10.83 ± 1.18^{a1b}	11.90 ± 1.42^{a2b}							^{b} = 0.044	^{b} = 0.820

[a]Denotes significant different between weeks 0 and 9 in each group.
[b]Denotes significant difference between the two groups at week 9.
TCHHC: traditional chinese herb hot compress.
TE: therapeutic exercise.

sichuan ox knee reduces the level of pro-inflammatory cytokines (TNF-α, IL-1β, IL-6, and IL-17A) in serum, increases the viscosity of intra-articular knee joint fluid, and eliminates swelling and promotes blood circulation (Weng et al., 2014; Li et al., 2017; Zhang et al., 2019a). Miyajima (Miyajima et al., 2013) et al. point out that hot compress increases the temperature of local tissues and reduces the excitability of afferent nerves, and helps control swelling and reduce pain. TCHHC allow herbs to penetrate the skin barrier through a warming effect, immediately reaching and acting on the affected areas. TCHHC can regulate the imbalance of the levels of anti-inflammatory factors (TGF-β1, IL-13), pro-inflammatory cytokines (IL-1β, IL-6), and pain mediators (PGE2, 5-HT), thus relieving pain (Orita et al., 2011). 2. TE facilitates TCHHC's pain-relieving effects, by enhancing TE enhances blood and lymphatic fluid circulation, as well as promoting local vasodilation (Guo et al., 2021); thus, it increases the absorption of herbs into the skin and promotes anti-inflammatory and pain-relieving effects.

At week 9, the proprioception thresholds of the knee extension and ankle plantar flexion decreased significantly in both groups, and the decrease was more significant in the TCHHC + TE group than in the TE group. Previous studies showed that TE has a positive effect on the proprioception of knee and ankle joints (Shen et al., 2021), which is consistent with our study. As the most important proprioceptor within the knee joint, the muscle spindles are stimulated by muscle lengthening, speed, or acceleration (Orita et al., 2011).TE releases and stretches the tensed muscles around the knee joint, coordinates the contraction ability between muscles, and enhances muscle spindle sensitivity to improve proprioception among older adults with KOA (Fransen et al., 2002). There are several possible reasons why TCHHC + TE had a superior effect on proprioception, as follows. 1. In this study, some of the components of the herbs have been proven to recover proprioception, such as cassia twig, mulberry parasitic, and antler gum prevent the loss of proteoglycans and accelerate the proliferation of chondrocytes to protect joint cartilage cells and improve their metabolism (Xu and Liu, 2004; Fan et al., 2012), reduce the damage to mechanoreceptors, promote neuromuscular control, and improve proprioception (Chen et al., 2016). Nagashima et al. (Nagashima et al., 2006) and Rusminingsih et al. (Rusminingsih et al., 2020) pointed out that hot compresses applied to the knee joint to dilate blood vessels, increases the excitability of motor neurons and recruits many motor neurons to participate in activities (Levine, 2007), facilitates sensory input. 2. TCHHC enhanced TE's effects on proprioception. By reducing the excitability of nerve endings and eliminating swelling, thereby allowing the adhesions and atrophy of the knee joint and its surrounding soft tissues to recover as much as possible (Zhang et al., 2015), which facilitates the sensitivity of muscle spindles (Wang et al., 2010; SH, 2012).

At week 9, the times of TUG and 20-m walk decreased significantly in both groups, and the decrease was more significant in the TCHHC + TE group than in the TE group. TCHHC + TE is more effective in improving functional performance, which is consistent with a previous study (Song, 2022). Another study has shown that herbs are used to enhance muscle strength and function performance (Sellami et al., 2018). The level of magnesium (Mg) (Zeng et al., 2015) and calcium (Ca^{+}) (Li et al., 2016) in the serum are significantly decreased among older adults with KOA, which reduces their functional performance, increases the risk of falls, and accelerates the deterioration of KOA (Veronese et al., 2017; Heffernan et al., 2020). Some of the Traditional Chinese Herbs, e.g., paeony root, ephedra root, atractylodes macrocephala reduce pro-inflammatory cytokines such as TNF- α and interleukins, inhibit the production of matrix metalloproteinases (Yuan et al., 2015; Malemud, 2017), thereby improve the level of Mg and Ca^{+} in serum (Gunn et al., 2013; Zhang et al., 2019a). In addition, parsnip and anemarrhena asphodeloides reduce the expression of fibroblast-like synoviocytes and receptor activator for NF-kB ligand, inhibit the activation of osteoclasts, and reduce the destruction of cartilage, bone, and tendon (Yuan et al., 2015). Previous studies have pointed out that hot compress promotes blood circulation, activates the motor cortex, and facilitates the recovery of adhesions and atrophy of the muscles and soft tissues around the knee joint (Zhang et al., 2015). Therefore, TCHHC is more effective in improving functional performance by expands capillaries, promotes blood circulation, and releases muscle tension (Savaş et al., 2019). The combination of TE facilitates herb penetration in the affected area, reduces stress-related muscle tension, improves lower extremity coordination, and enhances muscle strength to improve functional performance (Guo et al., 2021).

This study has several limitations. First of all, the effects of the TCHHC are possibly attributed to Chinese herbs or hot compress, further studies are recommended to further investigate their specific effects. Second, there was no follow-up after the 8-week intervention, and it was impossible to determine how long the effects would last.

5 Conclusion

Compared with TE, the 8-week TCHHC + TE was superior in relieving pain, recovering proprioception, and improving functional performance among older adults with KOA. It is recommended that TCHHC should be adopted prior to TE to enhance the effects of KOA rehabilitation.

Data availability statement

The datasets presented in this study can be found in online repositories. The names of the repository/repositories and accession number(s) can be found below: DOI:10.57760/sciencedb.02151.

Ethics statement

The studies involving human participants were reviewed and approved by the Ethics Committee of Neck, Shoulder, Back, and Leg Pain Hospital. The patients/participants provided their written informed consent to participate in this study.

Author contributions

JW designed the study and drafted the manuscript. WL was responsible for recruiting patients and collecting data. HF provided efforts in data analysis and revised the manuscript. All authors have read and agreed to the published version of the manuscript.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Relationship between proprioception and balance control among Chinese senior older adults

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Background: Balance impairment is the most common risk factor for falls among older adults, with three potential factors (tactile sensation, proprioception, and muscle strength) being responsible for their balance control. However, controversies remain on whether or not balance control is related to the three contributors among older adults. Therefore, clarifying the above questions helps explain the mechanisms of increased falls among senior older adults. This study compares the balance control and the three factors and investigates their relationships among older adults of different ages.

Methods: 166 participants ultimately passed the qualification assessment and were categorized into younger (YG, 60–69 years, n = 56), middle (MG, 70–79 years, n = 57), or older (OG, ≥80 years, n = 53) aged groups. Berg Balance Scale (BBS) performance, tactile sensation, proprioception, and muscle strength were tested. One-way ANOVA and partial correlation were performed to explore the differences between groups in BBS and its three potential contributors, along with the correlations between them within each age group.

Results: Significant differences among the three groups were detected in BBS scores ($p < 0.001$), tactile sensation at the great toe ($p = 0.015$) and heel ($p = 0.025$), proprioception of knee flexion ($p < 0.001$) and extension ($p < 0.001$), and ankle plantarflexion ($p < 0.001$) and dorsiflexion ($p < 0.001$), and muscle strength of ankle plantarflexion ($p < 0.001$) and dorsiflexion ($p < 0.001$), and hip abduction ($p < 0.001$). Proprioception of knee flexion ($r = -0.351$, $p = 0.009$) and extension ($r = -0.276$, $p = 0.041$), and ankle plantarflexion ($r = -0.283$, $p = 0.036$), and muscle strength of ankle plantarflexion ($r = 0.326$, $p = 0.015$) and hip abduction ($r = 0.415$, $p = 0.002$) were correlated with BBS among the YG. Proprioception of ankle plantarflexion ($r = -0.291$, $p = 0.030$) and muscle strength of ankle plantarflexion ($r = 0.448$, $p = 0.001$) and dorsiflexion ($r = 0.356$, $p = 0.007$) were correlated with BBS among the MG. Muscle strength of ankle plantarflexion ($r = 0.276$, $p = 0.039$) and hip abduction ($r = 0.324$, $p = 0.015$) were correlated with BBS among the OG.

Conclusion: YG and MG had better balance control, tactile sensation, proprioception, and muscle strength compared to OG. Proprioception correlated with balance control in YG and MG, but not in the OG. The worsen proprioception among the OG could be the key for increased falls.

Exercise should be recommended to improve proprioception among senior older adults.

KEYWORDS

balance control, tactile sensation, proprioception, muscle strength, older adult

1 Introduction

Falling is a major health concern for older adults and is one of the leading causes of injury and accidental death among older adults (Kannus et al., 2005; Pizzigalli et al., 2011). One study reported that accidental falls kill more than 10,000 people aged 65 and older yearly (Shankar et al., 2017) and this risk increases with age (Scuffham et al., 2003). Over 30% of adults aged over 60 years fall at least once a year (Moreland et al., 2003; Gerards et al., 2017) with the number increasing to 60% among those aged over 80 (Gschwind et al., 2013).

Impaired balance is one of the most common risk factors leading to falling (Muir et al., 2010). Therefore, performance-based functional balance tests have been widely adopted in laboratory tests to objectively quantify said risk. Specifically, the Berg Balance Scale (BBS) is commonly used to evaluate the balance control ability of older adults, thus helping predict the risk of falling.

Balance control is essential for performing most daily life activities and involves integrating information from the musculoskeletal systems and sensory inputs (Muir et al., 2010). Tactile sensation, proprioception, and muscle strength are three potentially responsible factors for balance control among older adults (Song et al., 2021). Tactile sensation plays an essential role in balance control, while cutaneous mechanoreceptors in the foot soles provide necessary feedback for the balance control system to maintain balance (Shaffer and Harrison, 2007). Proprioception is the internal sense of body position, and is also essential for regulating balance and generating and maintaining precise movement patterns or gaits (Henry and Baudry, 2019). Motion perception is important in maintaining balance among older adults: when a disturbance occurs, the lower extremity angles (such as the ankle angles) changes dramatically. Precepting the angle changes earlier would make it easier to resist the disturbance (Song et al., 2021). As the primary support joints of the human lower extremity, proprioception input around the knee and ankle provide a component for the establishment and maintenance of functional joint stability (Lephart et al., 1998). Muscle strength is then essential to maintain an upright posture and physical balance (Gouveia É et al., 2020). When individuals are subjected to external disturbances, they first restore their bodies to its original position by applying ankle or hip torques (Horak, 2006). Ankle plantarflexion and dorsiflexion and hip abduction muscle strength are particularly critical in preventing anterior-posterior and medial-lateral falls among older adults (LaRoche et al., 2010; Arvin et al., 2016).

Previous studies have long established the existence of age-related declines in balance control (Thomas et al., 2019), tactile sensation (Shaffer and Harrison, 2007), proprioception (Henry and Baudry, 2019), and muscle strength (Doherty, 2001). The decline of balance (Muir et al., 2010), sensation (Chen and Qu, 2019), and muscle strength (Yang et al., 2018) among older adults increases their risk of falling, yet few studies have investigated whether these factors continue to decline with age in older adults at different ages. Additionally, the relationship between tactile sensation (Menz et al., 2005; Ünver and Akbaş, 2018), proprioception (Amin and Herrington, 2014; Chen and Qu, 2019), and muscle strength (Muehlbauer et al., 2012; Gouveia É et al., 2020) with balance control remains controversial. These controversies may be due to the different relationships between balance control and potential factors in various age groups. For instance, Ünver and Akbaş. (2018) demonstrated no correlation between tactile sensation and balance among older adults with a mean age of 73, while Menz et al. (2005) showed that tactile sensation was related to balance among older adults with an average age of 80 years. To our knowledge, studies are yet to explore the relationship between balance control and its potential factors among older adults in different age groups.

Therefore, this study pioneeringly compares balance control, tactile sensation, proprioception, and muscle strength across three different age groups and examines their age-specific relationships to determine the causes of increased falls among older adults. Our primary hypothesis was significant differences in balance control, tactile sensation, proprioceptive, and muscle strength among older adults of different ages. The secondary hypothesis posits that balance control was significantly correlated with tactile sensation, proprioceptive, and muscle strength among older adults from different age group.

2 Material and methods

2.1 Participants

Participants were recruited through awareness talks and the distribution of flyers in the local community. The inclusion criteria were as follows: 1) ≥ 60 years; and 2) walking independently without outside help. Meanwhile, the exclusion criteria were the following: 1) visual defects, vestibular disorders, central nervous system dysfunction, and psychological disorders associated with previous falls; 2) cognitive impairment detected through laboratory

examination; and 3) self-reported trauma or disease that might significantly affect BBS, tactile sensation, proprioception and muscle strength testing such as chronic pain, history of fracture during the last 6 months, foot ulcers, osteoarthritis, diabetes, lower extremity joints prostheses and lumbar disc herniation. Following the qualification assessment, a total of 166 participants were enrolled and included in the final analysis. The age groups of the older adults were identified as younger- (YG, 60–69 years, n = 56), middle- (MG, 70–79 years, n = 57), or older- (OG, ≥80 years, n = 53) age groups. All participants voluntarily signed informed consent forms before the start of the test. All protocols and procedures herein were approved by the Institutional Review Boards in Shandong Sport University (19003) and followed the Helsinki Declaration.

2.2 BBS test

The BBS test was used for the functional assessment of balance control, which included 14 tests of simple daily functional activities such as standing with eyes closed, single leg stance, and sit-to-stand conversion (ICC value, 0.98–0.99) (Berg et al., 1992). The participants' performance in each of these activities was measured by a five-point ordinal scale ranging from 0 to 4 (0 = could not complete the task; 4 = could complete the task independently), making a possible total ranged of 0–56, with higher scores indicating better balance (Downs et al., 2014). During evaluation, researchers evaluated and scored participants by observing their performance.

2.3 Tactile sensation tests

The tactile sensitivity of the dominant foot was tested using a set of Semmes-Weinstein monofilaments (North Coast Medical, Inc., Morgan Hill, CA, United States; ICC value, 0.83–0.86) (Collins et al., 2010). The six sizes of monofilaments used herein were 2.83, 3.61, 4.31, 4.56, 5.07, and 6.65, respectively. Pressure was applied until the monofilament formed a C-shaped bend (bend 90°), where the applied force values were 0.07, 0.4, 2, 4, 10, and 300 g. Then, participants were placed in a supine position on the treatment table and randomly tested for tactile sensitivity in the heel, arch, first and fifth metatarsals, and big toe. Thinner monofilaments were used at the beginning of the test and were gradually increased to thicker monofilaments until the participant was able to detect thorough their touch. The sensitivity threshold was determined by the minimum monofilament gauge detected correctly (Ünver and Akbaş, 2018). Better tactile sensitivity was indicated by the lower sensitivity threshold with the lowest threshold detected recorded for data analysis.

2.4 Proprioception tests

The proprioceptive thresholds were assessed using a proprioceptive tester, which showed good test-retest reliability (Toshimi, Jinan, Shandong, China; ICC value, 0.737–0.935) (Sun et al., 2015). Proprioceptive thresholds were collected during ankle dorsi/plantarflexion and knee flexion/extension. The apparatus consisted mainly of an operating panel, a steel frame for suspension of the lower limbs, and two movable pedals. At the beginning of the test, the researcher turned on the movable pedal with a button on the control panel. During both ankle and knee proprioceptive tests, the shank was in slight contact with the pedal surface and perpendicular to the pedal surface, while the knee and hip joints were flexed at a 90° angle. Participants were seated in a test chair and wore eye masks and headphones playing music to eliminate potential environmental visual and auditory stimuli while asking the participants to focus their attention. As soon as participants could feel the movement, they were required to press a hand switch to cease the movement of the platform and subsequently identify the direction of the movement. All participants wore socks with uniform material and thickness for the proprioception test. The test was considered successful if participants correctly identified the direction of motion. Three trials were used in each direction for the final data analysis.

2.5 Muscle strength tests

Strength testing was assessed using the IsoMed 2000 strength testing system (D. & R. Ferstl GmbH, Hemau, Germany; ICC value, 0.77–0.98) (Gonosova et al., 2018). During the ankle muscle strength test, participants were placed in the supine position with the hip and knee joints in full extension. Ankle motion ranged from 5° of dorsiflexion to 30° of plantarflexion. Meanwhile, in the hip strength test, participants were placed in a lateral position with the hip fully extended, the non-test leg slightly flexed, and the test leg with the knee extended. Additionally, the participant was secured to the bed by a belt around the pelvis and the non-test leg to maintain stability. Moreover, the test leg was stabilized by the belt and the participant was asked to abduct the tested leg with maximum force with a range of motion from 0° to 30°. Participants were then asked to perform a maximum isokinetic force test at an angular velocity of 10°/s and a break was given of at least 2 min between two consecutive trials. Three tests were conducted in each direction.

2.6 Statistics

SPSS 26.0 was used to perform the statistical analysis. The Shapiro-Wilk test was then used to check the normality of the

TABLE 1 Participants' information.

Group	YG	MG	OG	*p*
	(*n* = 56, female = 29)	(*n* = 57, female = 28)	(*n* = 53 female = 36)	
Age (years)	65.45 ± 2.15	74.86 ± 3.11	85.21 ± 2.74	**<0.001**
Height (m)	1.64 ± 0.08	1.62 ± 0.07	1.60 ± 0.07	**0.014**
Weight (kg)	65.78 ± 10.94	65.07 ± 9.26	62.81 ± 7.83	0.241
BMI (kg/m^2)	24.39 ± 3.18	24.72 ± 3.23	24.56 ± 2.73	0.855

Presented as mean ± standard deviation. YG, The younger-aged groups (60–69 years); MG, The middle-aged groups (70–79 years); OG, The older-aged groups (≥80 years); BMI: Body mass index. Bold: $p < 0.05$.

TABLE 2 Descriptive characteristics of muscle strength, tactile sensation, and proprioception.

Variables		YG	MG	OG	*p*
BBS		54.93 ± 1.18[a, b]	53.46 ± 1.93[c]	50.25 ± 2.66	**<0.001**
Tactile sensation (gauge)	Great toe	4.18 ± 0.53[a, b]	4.35 ± 0.56	4.55 ± 0.81	**0.015**
	1st Metatarsal	4.26 ± 0.61	4.24 ± 0.59	4.32 ± 0.48	0.112
	5th Metatarsal	4.36 ± 0.46	4.34 ± 0.48	4.43 ± 0.48	0.386
	Arch	4.41 ± 0.40	4.47 ± 0.64	4.47 ± 0.51	0.492
	Heel	4.51 ± 0.50[a]	4.65 ± 0.57	4.73 ± 0.66	**0.025**
Proprioception (°)	Knee flexion	2.25 ± 1.41[a, b]	2.91 ± 1.95[c]	3.86 ± 2.35	**<0.001**
	Knee extension	2.46 ± 1.39[a, b]	3.33 ± 2.60[c]	4.05 ± 2.62	**<0.001**
	Ankle plantarflexion	1.99 ± 1.05[a, b]	3.21 ± 2.59	5.88 ± 4.38	**<0.001**
	Ankle dorsiflexion	2.29 ± 1.76[a, b]	3.42 ± 3.18[c]	5.86 ± 4.00	**<0.001**
Strength (N*m/kg)	Ankle plantarflexion	0.46 ± 0.17[a, b]	0.35 ± 0.13[c]	0.23 ± 0.13	**<0.001**
	Ankle dorsiflexion	0.25 ± 0.06[b]	0.22 ± 0.07[c]	0.17 ± 0.07	**<0.001**
	Hip abduction	0.51 ± 0.16[a, b]	0.41 ± 0.16[c]	0.32 ± 0.15	**<0.001**

Presented as mean ± standard deviation. YG, The younger-aged groups (60–69 years); MG, The middle-aged groups (70–79 years); OG, The older-aged groups (≥80 years); Bold: $p < 0.05$.
[a]Between-group differences of younger- and medium-older adults.
[b]Between-group differences of younger- and oldest-older adults.
[c]Between-group differences of medium- and oldest-older adults.

data distribution. The means and standard deviations of all outcome variables were descriptively analyzed. Meanwhile, one-way ANOVA (normality) or Kruskal-Wallis H tests (non-normality) were used to compare differences between aging groups. Post hoc analysis was performed using Bonferroni when there were significant differences between groups. Furthermore, Pearson (normality) or Spearman (non-normality) correlations were used to determine the relationship of BBS with tactile sensation, proprioception, and muscle strength in each group. Participants' heights were also adjusted as covariates. The thresholds of correlation coefficient (r) were as follows: >0.5 (strong); 0.5–0.3 (moderate); 0.3–0.1 (weak); 0.1–0 (trivial) (Cohen, 1988).

3 Results

All variables except BBS, tactile sensation, and proprioception were normally distributed. A one-way ANOVA showed significant differences in age ($p < 0.001$) and height ($p = 0.014$) among the three groups. However, no statistical differences were detected in weight and body mass index (BMI) among the three groups (Table 1).

The descriptive statistics of BBS, tactile sensation, proprioception, and muscle strength are shown in Table 2. Results revealed significant differences in BBS scores, tactile sensation, proprioceptive, and muscle strength across the three age groups. A higher score of BBS was observed in the YG compared to the MG ($p = 0.001$) and OG ($p < 0.001$). The

TABLE 3 The age-specific correlations of BBS with muscle strength, tactile sensation, and proprioception.

Variables		BBS					
		YG		MG		OG	
		r	*p*	*r*	*p*	*r*	*p*
Tactile sensation (gauge)	Great toe	−0.101	0.461	−0.213	0.115	−0.264	0.051
	1st Metatarsal	0.047	0.732	−0.117	0.391	−0.016	0.909
	5th Metatarsal	0.067	0.625	−0.169	0.209	0.089	0.513
	Arch	−0.045	0.745	−0.256	0.056	0.062	0.648
	Heel	−0.129	0.347	−0.021	0.878	−0.004	0.975
Proprioception (°)	Knee flexion	−0.351	**0.009**	−0.241	0.073	−0.126	0.354
	Knee extension	−0.276	**0.041**	−0.060	0.662	−0.101	0.460
	Ankle plantarflexion	−0.283	**0.036**	−0.291	**0.030**	−0.134	0.324
	Ankle dorsiflexion	−0.196	0.151	−0.207	0.125	−0.192	0.257
Strength (N*m/kg)	Ankle plantarflexion	0.326	**0.015**	0.448	**0.001**	0.276	**0.039**
	Ankle dorsiflexion	0.126	0.360	0.356	**0.007**	0.096	0.480
	Hip abduction	0.415	**0.002**	0.244	0.071	0.324	**0.015**

YG, The younger-aged groups (60–69 years); MG, The middle-aged groups (70–79 years); OG, The older-aged groups (≥80 years); *r*: correlation coefficient. Bold: $p < 05$. Adjusted for height.

MG had worse tactile sensation in the great toe ($p = 0.035$) and heel ($p = 0.047$), had higher proprioception threshold of knee flexion ($p = 0.030$) and extension ($p = 0.030$), and ankle plantarflexion ($p = 0.010$) and dorsiflexion ($p = 0.039$), and had less muscle strength in ankle plantarflexion ($p = 0.010$) and hip abduction ($p = 0.002$) compared to the YG. Meanwhile, the OG had worse tactile sensation in the great toe ($p = 0.005$), higher proprioception threshold of knee flexion ($p < 0.001$) and extension ($p < 0.001$), ankle plantarflexion ($p < 0.001$) and dorsiflexion ($p < 0.001$), and had less muscle strength in ankle plantarflexion ($p < 0.001$) and dorsiflexion ($p < 0.001$), and hip abduction ($p < 0.001$) compared to the YG. They also had higher proprioception threshold of knee flexion ($p = 0.021$) and extension ($p < 0.001$), ankle dorsiflexion ($p = 0.001$), and less muscle strength in ankle plantarflexion ($p < 0.001$) and dorsiflexion ($p = 0.001$), and hip abduction ($p = 0.010$) compared to the MG.

The correlations of BBS with tactile sensation, proprioception, and muscle strength are shown in Table 3. Among the YG, the BBS was weakly to moderately correlated with proprioception of knee flexion ($r = −0.351$, $p = 0.009$) and extension ($r = −0.276$, $p = 0.041$), and ankle plantarflexion ($r = −0.283$, $p = 0.036$). The BBS was moderately to strongly correlated with muscle strength of ankle plantarflexion ($r = 0.326$, $p = 0.015$) and hip abduction ($r = 0.415$, $p = 0.002$). However, none of the BBS was correlated with tactile sensation test results. Among the MG, the BBS was weakly correlated with proprioception of ankle plantarflexion ($r = −0.291$, $p = 0.030$) and was moderately correlated with muscle strength of ankle plantarflexion ($r = 0.448$, $p = 0.001$) and dorsiflexion ($r = 0.356$, $p = 0.007$). However, none of the BBS was correlated with tactile sensation test results. Among the OG, the BBS was weakly to moderately correlated with muscle strength of ankle plantarflexion ($r = 0.276$, $p = 0.034$) and hip abduction ($r = 0.324$, $p = 0.015$), with none of the BBS still being correlated with tactile sensation and proprioception test results.

4 Discussion

This study compared balance control and its potential contributors, tactile sensation, proprioception, and muscle strength and investigated the relationship of balance control with its three potential contributors among older adults of different ages. Results partly support our hypotheses: there were significant differences observed in balance control, tactile sensation, proprioception, and muscle strength across the three groups. The balance control was correlated with proprioception and muscle strength in the YG and MG, but it was only correlated with strength in the OG. None of the tactile sensation was correlated with balance control in the three groups.

A lower score of BBS was observed in the OG compared to the other two groups. The findings were consistent with previous studies that a decline in the balance functions occurs with age

(Cuevas-Trisan, 2017; Thomas et al., 2019). Therefore, the balance function is essential to prevent falling in older adults. Changes in the central nervous system and neuromuscular system properties with age may negatively affect the performance of balance control among older adults (Gschwind et al., 2013).

The outcomes have shown that the OG had worse tactile sensation compared to the MG and YG, and no correlations were confirmed between tactile sensation and balance control among the three age groups. Our findings are consistent with those of some previous reports (Shaffer and Harrison, 2007). Aging may affect the mechanical properties of the skin as well as changes in skin receptor density, morphology, and physiology, likely leading to a reduced tactile sensation among older adults (Peters et al., 2016). The great toe and heel are located at the front and rear of the foot and at the most distal part of the body, while the tactile sensation deteriorates in a distal to proximal manner (Toledo and Barela, 2014). This may explain the significant group differences at both the great toe and heel. Further, the tactile sensation was not correlated with balance control, which has been documented in previous studies (Ünver and Akbaş, 2018). However, some studies reported that tactile sensation was correlated with balance control (Bretan et al., 2010). This conflict may be explained by the difference in monofilament specifications used in the different experiments. Bretan et al. (2010) used only one monofilament sized 5.07 (10 g), while six monofilament sizes were used to test tactile sensation in the current study which allowed for more comprehensive data. By using six monofilaments of different sizes, tactile sensation threshold of the older adult could be measured more accurately. Moreover, the lack of correlation between tactile sensation and balance control among older adults may also be due to the compensation of other sensory information for tactile sensation when maintaining balance such as proprioceptive, visual, and vestibular sensory information (Chen et al., 2012; Ferlinc et al., 2019).

Results showed that OG had a higher proprioception threshold compared to the other two groups, and proprioception was related to balance control in YG and MG, but not in the OG. Proprioception therefore declines with age, which has been well documented in existing studies (Henry and Baudry, 2019). The decline of proprioception may be related to age-related alterations in the musculus and its neural pathways, which may lead to deficits in the processing and input of proprioceptive signals (Ferlinc et al., 2019). It may also be associated with muscle dysfunction and degeneration of articular cartilage (Wingert et al., 2014). Here, joint motion sense was used to represent proprioception, rather than joint position sense. Notably, the participants' performance on joint position sense was overall more erratic compared with joint motion sense (error varied more among trials) (Reider et al., 2003). Joint position sense is a more complex test than joint motion sense, which requires coordinated afferent and supraspinal efferent output. In joint motion sense, subjects only have to signal when motion is sensed. In joint position sense however, the subjects must try to remember a position and then accurately reproduce it. Thus, there may have been more of a study effect in joint position sense findings with the older adults who were tested on multiple occasions. They may have tended to concentrate more than the external controls and may also have benefited from a learning effect (Reider et al., 2003). Moreover, joint force sensing was not used because it did not show good reliability (Benjaminse et al., 2009) due to the difficulty for participants to maintain a uniform test posture during the test and because participants may use different strategies to complete the test (Benjaminse et al., 2009). There was no significant difference in the proprioception of ankle plantarflexion between the MG and OG groups. This could be due to the larger standard deviation in the OG group. Sensation deteriorates in a distal to proximal manner (Toledo and Barela, 2014), and more distal muscles, such as the soleus, which is one of the ankle plantarflexors, may deteriorate more than other muscles and enlarge the variance in plantarflexion proprioception in the OG group. Knee proprioception was not correlated with BBS scores in the MG and OG groups, which may be due to the ankle or hip strategies being often used to return the body to balance when disturbed instead of the knee strategy (Horak, 2006). The proprioceptive feedback in the ankle joint may also be more important in regulating muscle activity compared to the knee joint (Mayer et al., 2018). Our findings indicated that dorsiflexion proprioception is not correlated with BBS scores, while plantarflexion proprioception is correlated with BBS scores. This may be because individuals rely more on the contraction of the plantarflexor (rather than the dorsiflexor) to complete most of the BBS tasks, such as from sitting to standing or when turning. Previous studies coincide with this and point out that ankle proprioception is crucial for balance and relies upon accurate input from plantarflexor, such as the calf triceps (Reynolds et al., 2020). Meanwhile, proprioception was correlated with balance control in both YG and MG groups, but not among the OG. Proprioception is the sensory system that provides information about motor activity levels to the central nervous system, and its inputs are essential for balance control (Horak, 2006). However, the decline in proprioception may significantly increase body sway and affect mobility, thereby resulting in impaired balance and a higher possibility of falling (Ferlinc et al., 2019). Therefore, proprioception was related to balance control in the YG and MG groups—older adults in these two groups could thus rely on proprioception to control balance. However, proprioception was not related to balance control in OG, which may be that the proprioception among the OG could not provide sufficient information on balance control. This inferred the deterioration of proprioception (to a certain extent) among older adults over 80 years, and did not continue to provide any meaningful

information on balance control in older adults. Furthermore, proprioception deterioration may have a greater effect on balance control among older adults older than 80, which may also plausibly account for the increased risk of falling of those in the OG.

Results herein have shown that weaker muscle strength was observed in the OG compared to the other two groups, and that muscle strength was related to balance control among the three age groups. This coincides with the outcomes of other studies where aging is associated with a significant decline in muscle strength (Doherty, 2001). Additionally, balance control was correlated with muscle strength among older adults, which is also consistent with previous studies (Tavakkoli et al., 2021). Similarly, Tavakkoli et al. (2021) found a significant correlation between muscle strength of ankle plantarflexion with balance among older adults. These observations also contrast those of other studies. Muehlbauer et al. (2012) found no significant correlation between muscle strength and balance among older adults. The inconsistent results may be because their study measured muscle strength during isometric contractions, whereas the current study measured muscle strength during isokinetic contractions (Muehlbauer et al., 2012). Isokinetic contractions involve dynamic muscle contraction and can better reflect dynamic balance control, compared to isometric contractions which involve static contractions (Song et al., 2021). Although muscle strength decreases with age, the relationship between muscle strength with balance control in all age groups may indicate that muscle strength still plays an important role in maintaining balance control among older adults.

The following limitations are seen herein: first, only the effects of tactile sensation, proprioception, and muscle strength on balance control were investigated in the current study. Nevertheless, other factors might also affect balance control among older adults such as visual, cognitive functions, or vestibular factors. Second, all the participants were recruited from the same city and had similar backgrounds. Therefore, it is recommended that future studies should include populations with different characteristics.

5 Conclusion

Senior older adults aged over 80 have worse balance control, deteriorated tactile sensation and proprioception, and less muscle strength compared to their younger counterparts. Proprioception and muscle strength were correlated with balance control in the 60–69 and 70–79 age groups, while only strength, but not proprioception, was correlated with balance control among older adults over 80. The worsened proprioception among older adults aged over 80 years could be the key for their increased falling.

Data availability statement

The datasets presented in this study can be found in online repositories. The names of the repository/repositories and accession number(s) can be found below: http://doi.org/10.57760/sciencedb.03500.

Ethics statement

The studies involving human participants were reviewed and approved by the Institutional Review Boards in Shandong Sport University (19003). The patients/participants provided their written informed consent to participate in this study.

Author contributions

All authors listed have made a substantial, direct, and intellectual contribution to the work and approved it for publication.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Characteristics of the postural stability of the lower limb in different visual states of undergraduate students with moderate myopia

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Objective: To explore the characteristics of lower limb postural stability in undergraduates with moderate myopia in three different visual states.

Methods: Twenty male undergraduate students were recruited to complete respectively the static and dynamic postural stability tests under eyes-closed, myopia (taking off their glasses immediately) and corrected vision conditions. A three-dimensional force platform (Bertec, United States) was used to test static postural stability, which calculated the total path length of the Center of Pressure (COP), path length in the antero-posterior (A/P) and medio-lateral (M/L) directions, COP area, SampleEntropy (SampEn), and low-, medium-, and high-frequency spectrum energies. Dynamic postural stability was tested using the Y-balance test, and the Y-balance test scores were calculated. The Vicon three-dimensional motion capture system (Oxford, United Kingdom) measured the maximum flexion angles of the ankle, knee, and hip joints. The electromyography (EMG) root mean square (RMS) and integral EMG (iEMG) of the tibialis anterior and lateral gastrocnemius of the lower extremity were simultaneously measured using wireless surface electromyography (Noraxon, United States).

Results: The SampEn-A/P and SampEn-M/L of corrected vision state higher than myopia and eyes-closed states, and myopia state larger than eyes-closed state (χ^2 = 51.631, p < .001). The original and standard scores of the anterior, postero-medial and comprehensive values of the three visual states had significant differences (F = 32.125, p < .001). The original and standard values of postero-lateral corrected vision and myopia were larger than those of eyes-closed states (F = 37.972, p < .001). The maximum flexion angles of the ankle and knee joints were in the following order: corrected vision, myopia and eyes-closed (F = 10.93, p < .001). The iEMG and RMS had significant differences in the three different states (χ^2 = 12.700, p < .001) in the all directions of YBT.

Conclusion: Compared with corrected vision, the stability of static posture in the state of myopia was decreased, and the postural regularity was more regular. The dynamic postural stability in the state of myopia was also lower than that corrected vision, and the activation and work of ankle muscles were also increased.

KEYWORDS

postural stability, moderate myopia, y-balance test, attention, EMG

1 Introduction

Postural stability refers to the ability to keep the body's center of pressure stable within the supporting area (Elboim-Gabyzon et al., 2022). The sensory inputs that are required to maintain postural stability in humans include vision, vestibular, and proprioception. When the visual information is blocked, the static postural stability swing of the human body will increase by 20%–70% (Isotalo et al., 2004). According to a recent prediction, the global population of people with myopia is expected to reach 5 billion by 2050 (Holden et al., 2016). Since the outbreak of the COVID-19 pandemic, the incidence and severity of myopia among undergraduate students appear to be on the increase (Fang and Guo., 2021). If myopia is not controlled promptly, it could easily progress to high myopia, which leads to a series of pathological changes, such as retinal detachment, glaucoma, cataract, and even permanent visual impairment or blindness (Lord and Dayhew., 2001). When myopia progresses to a certain extent, it caused blurring of vision, decreased contrast sensitivity and stereoscopic sense, and decreased postural stability (Ikuno, 2017). However, there are currently few studies on the effects of myopia on postural stability, and existing studies have focused on the ability to maintain static balance. According to Willis et al. (2013) found that people with myopia had more static postural instability than those with normal vision. Bae et al. (2020) found that the static postural stability of people with uncorrected vision (without glasses) was significantly lower than that of those with corrected vision. Myopia was also significantly higher than corrected visual acuity in the middle- and high-frequency spectral energy of the somatic system. Sayah et al. (2016) found that people with myopia were more unstable than those with emmetropia in terms of visually induced postural responses. The postural stability includes both static and dynamic aspects, and testing static stability alone cannot fully explain the characteristics of postural stability in people with myopia. The risk of falls in humans is influenced to a greater extent by dynamic postural stability than by static postural stability, and most falls occur during dynamic motion (Ray et al., 2008; Jeter et al., 2015). Due to the lack of accurate visual input and feedback after visual impairment, accidental injuries including falls, are more likely to occur during dynamic motion. Most studies have focused on epidemiology and its influencing factors (Liu et al., 2022; Pan, 2022).

When vision is impaired, the sensory input of visually impaired individuals appears as the vestibular and proprioception compensative mechanism (Schwesig et al., 2011). Labanca et al. (2021) compared with eyes-opened, the ankle muscle group co-activation and ankle strategy of young people were more obvious when the posture was unstable, the neuromuscular system could improve the stability by regulating the activity of the ankle muscle group in the eyes-closed state. However, it is unclear whether myopia affects muscle activation.

Therefore, the objective of this study was to explore the characteristics of static and dynamic postural stability in undergraduate students with moderate myopia in eyes-closed, myopia, and corrected vision states. In this research would assume: 1) The stability of static posture in the myopia state was lower than that in the corrected vision state, which was better than that in the eyes-closed state. 2) The stability of dynamic posture in the myopia state was lower than that in the corrected vision state, which was better than that in the eyes-closed state. 3) In the dynamic stability test, the activation intensities of the tibialis anterior (TA) and gastrocnemius (GA) were in the following order: eyes-closed, myopia, and corrected vision states.

2 Materials and methods

2.1 Participants

From April to June 2022, 20 male undergraduate students with moderate myopia (age: 19.5 ± 1.10 years; height: 1.75 ± .06 m; weight: 72.64 ± 13.63 kg; visual acuity: 4.33 ± −80 D (based on a verbal declaration and used the International Standard Logarithmic Visual Acuity Chart (Chen et al., 2020); and lower limb length: .90 ± .56 m) in Binzhou Medical University were recruited.

The inclusion criteria were as follows: 1) the age was >16 years; 2) the corrected vision was normal, and the better visual acuity was moderate myopia (-3.0D to -6.0D) (Goes and Delbeke., 2022); 3) non-sports major students; and 4) participants who signed informed consent forms.

The exclusion criteria were as follows: 1) abnormal musculoskeletal function, or proprioception, vestibular function, disturbance of consciousness, severe cardiovascular disease, *etc.*; 2) history of neuropathy or administration of drugs that may affect balance; 3) damage to the inner ear and another sensory system that may affect balance; and 4) difference in astigmatism, strabismus, and binocular vision >100°; 5) those who have received balance training or regular physical activity (regular physical activity was defined as undergoing vigorous-intensity physical activity for >20 min or moderate-intensity physical activity for >30 min, at least once weekly (Park et al., 2022) in the past 6 months.

The study followed the tenets of the Declaration of Helsinki and was approved by the ethical review board of the scientific research project of the Medical Ethics Committee of Binzhou Medical University (Ethics approval No: 2021–233).

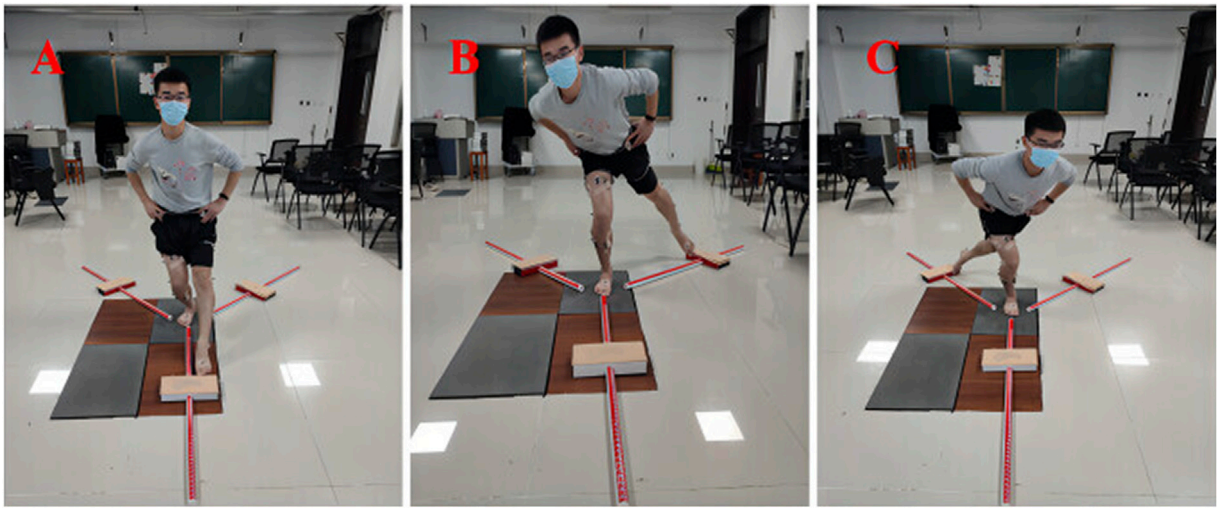

FIGURE 1
The picture of the subject during the YBT. **(A)** The anterior direction of the YBT. **(B)** The postero-medial direction of the YBT. **(C)** The postero-lateral direction of the YBT.

2.2 Procedure

2.2.1 Static postural stability test

The static postural stability was tested using a three-dimensional force platform (Bertec, United States) with a sample rate of 1000 Hz to collect static postural stability data (Gómez-Landero et al., 2021). Before the test, the participants performed warm-up exercises. The participants stood barefoot in the center of the force platform, while looking forward with the distance between the feet being equal to their shoulder width, and their upper limbs hung naturally on both sides of the body, and they remained motionless for 25 s. The data were taken from the middle 15 s, and the first 5 s and last 5 s were removed (Pryhoda et al., 2020). Measurements were taken three times under eyes-closed, myopia (tested immediately after taking off the glasses), and corrected vision states, and testing in this order. The average values were calculated. After each measurement, the participants rested for at least 1 min.

The indicators for evaluating the static postural stability include the linear indicators (the total path length of the Center of Pressure (COP), et al.), the non-linear indicators (SampleEntropy (SampEn)), and frequency spectral energy (low-, middle-, and high-frequency spectral energy). SampEn is the statistical regularity of the time series, with low (high) values indicating more (less) regularity (Stins et al., 2009). It is generally believed that the lower the SampEn, the more regular the COP signal regularity. The lower the automation of postural control, the higher the requirement for attention (Roerdink et al., 2011; Kędziorek and Błażkiewicz., 2020). Postural stability is maintained through the integration of sensory information provided by the visual, proprioceptive, and vestibular systems (Loughlin and Redfern., 2001). Those systems have different time delays in their control pathways, enabling their relative afferent contributions to be studied by identifying characteristic COP frequency responses (Daniels et al., 2019). The frequency spectral energy can be obtained by performing a fast Fourier transform, which enables the researcher to convert wave graphs of COP movement as represented by amplitude (power) and time into a relationship between amplitude (power) and frequency (spectrum); the power included for each frequency is defined as the frequency spectral energy (Tanaka et al., 2020). It is generally agreed that visual information of postural sway at low frequency (0–.3 Hz), vestibular information of postural sway at intermediate frequency (.3–1 Hz), and proprioceptive information of postural sway at high frequency (1–3 Hz) (Kanekar et al., 2014).

2.2.2 Dynamic postural stability test

The dynamic postural stability test used the Y-balance test (YBT) exercise kit to collect dynamic postural stability data (Salas-Gómez et al., 2022). The YBT includes the anterior A) (Figure 1A), postero-medial (PM) (Figure 1B), and postero-lateral (PL) (Figure 1C) measurements. During the YBT, the VICON MX motion analysis system (Oxford, United Kingdom), with a sample rate of 100 Hz (Tang et al., 2022) was used to record the maximum flexion angles of the ankle, knee, and hip joints. Surface electromyography (Noraxon, United States) with a sample rate of 1500 Hz (Yu et al., 2022) was used to test the EMG signal characteristics of the TA and lateral GM. Recording the EMG signal from the beginning of the extension of the leg to the extension to the farthest.

The disposable surface electrodes for EMG was attached to the participants after the warm-up exercises. First, the body hair was removed, the skin was disinfected with 75% alcohol, after which the skin was wiped repeatedly to reduce impedance. The EMG was then attached to the belly of the muscles along the direction of the muscle fibers after the skin was dry, and the sensor was connected and fixed. The Maximum Voluntary Contraction (MVC) was tested using the manual resistance methods (Balasukumaran et al., 2020). The muscle group around the ankle of the dominant leg was tested twice for 5 s each time at intervals of 10 s. The average value of the Root Mean Square (RMS) in the two MVC tests of each muscle was used to standardize the RMS (Mueller et al., 2018). The landmarks of the Vicon motion capture system were modeled according to the Plug-in-Gait of the lower body, and 16 marker balls were affixed to the subjects' landmarks.

First, the dominant leg (the leg the participants first choose to kick a ball (van Melick et al., 2017) was determined. While the participants stood on the dominant leg, the non-dominant leg was used as the forward extension leg. The participants stood barefoot in the center of the YBT exercise suite, and the dominant toe was aligned directly in front of the central horizontal line, while the participants held their waist with both hands. The right leg was tested counterclockwise, while the left leg was tested clockwise. The next direction was changed after three measurements in each direction. All measurements had an accuracy of ±.5 cm (Zafar et al., 2020). Before the formal test, the participants were asked to make six attempts in each direction of each leg to familiarize themselves with the test movement. To avoid the influence of learning, the participants were required to take a 5-min break before taking the formal test. If the participants failed to stretch out in six attempts, the direction was marked as 0 cm.

Measurements were taken three times under the eyes-closed, myopia (tested immediately after taking off the glasses), and corrected vision states, and testing in this order. The average values were calculated. After each measurement, the participants rested for 1 min.

Note: The following failures were not included in the data analysis: loss of balance while standing on one leg, obvious movement of the standing foot, landing support of the extended foot, and failure to return to the original position.

2.2.3 Evaluation index

The linear indicators were: the total path length of the COP, path length in the antero-posterior (A/P) and medio-lateral (M/L) directions, and the COP area were measured. The non-linear indicators were: SampEn, low-, middle-, and high-frequency spectral energy. The original and standard values of the extension distances in the three directions of the YBT (the comprehensive value of the extension distance; the maximum angles of the ankle, knee, and hip joint movement; RMS; and integral EMG value (iEMG)) were also evaluated.

2.3 Data process and analysis

The EMG signals were smoothly filtered using Noraxon MR3 software, and were rectified along with band-pass filtering (10–500 Hz) (Coratella et al., 2021). Finally, COP and EMG signal data were calculated and processed using MatLab 2020 software (The Math Works, Natick, MA, United States).

2.4 Statistical analysis

A priori power analysis was performed with G*Power 3.1.9.7 software. Using the Repeated Measures Analysis, setting the alpha error at .05, the Power at .80, and comparing three states in one group with an effect size (f = .50), showed that a minimum of 15 participants. Finally, we recruited 20 participants. The statistical calculations were carried out using SPSS 22.0 (IBM, Armonk, NY, United States) software for statistical analysis of the data. The normality of each data set was evaluated by the Shapiro-Wilk normality test. Quantitative data that were normally distributed were expressed as mean ± standard deviation, and F-tests were used to determine differences in the three states, i.e., the scores of YBT and angles of joints. This experiment was a compatibility group experiment to explore the differences between the same sample under three different visual states. It was a randomized block design and also belongs to the related sample design. Repeated measurement analysis of variance was not in line with normality, expressed by Median values (25 percent of quartiles, 75 percent of quartiles), using non-parametric Friedman's test. Chi-squared tests were used to determine differences in the three states, i.e., the index of COP, RMS, and iEMG. A Bonferroni correction factor was applied if significant. The criterion of statistical signification was set at .05.

3 Results

The linear indexes showed corrected vision lower than eyes-closed state ($p < .001$) (Table 1). There were no differences between eyes-closed and myopia state or between myopia and corrected vision state ($p > .05$) (Table 1). The sample entropy of non-linear indicators showed that visual factors had a significant effect on postural regularity ($p < .001$) (Table 1). The middle- and high-frequency spectrum energies during eyes-closed and myopia states were higher than those of corrected vision states ($p < .001$) (Table 1).

TABLE 1 Comparison of static postural stability of lower limbs in different visual states (Median values (25 percent of quartiles, 75 percent of quartiles)).

Project	Eyes-closed	Myopia	Corrected vision	χ^2-value	*p*-value	Post-hoc comparisons		
						Eyes-closed VS. myopia	Eyes-closed VS. corrected vision	Myopia VS. corrected vision
						p-value	*p*-value	*p*-value
Total path length of COP(m)	0.24 (0.20, 0.29)	0.21 (0.15, 0.27)	0.21 (0.16, 0.24)	12.100	0.002	—	0.002	—
Path length of $COP_{A/P}$(m)	0.18 (0.13, 0.23)	0.16 (0.12, 0.21)	0.15 (0.12, 0.19)	5.853	0.007	—	0.020	—
Path length of $COP_{M/L}$(m)	0.11 (0.09, 0.14)	0.10 (0.07, 0.14)	0.09 (0.07, 0.13)	11.100	0.004	—	0.003	—
COP area (cm^2)	616.25 (286.96, 1360.10)	577.50 (175.12, 1278.00)	488.87 (192.26, 890.41)	7.900	0.019	—	0.022	—
SampEn-A/P	0.85 (0.58, 0.99)	1.02 (0.85, 1.26)	1.38 (1.17, 1.61)	51.631	<0.001	<0.001	<0.001	<0.001
SampEn-M/L	0.76 (0.55, 0.94)	1.02 (0.82, 1.22)	1.24 (.93, 1.43)	76.960	<0.001	<0.001	<0.001	<0.001
Low spectrum energy (%)	67.56 (54.95, 74.38)	72.81 (63.25, 81.13)	81.51 (69.41, 84.30)	23.070	<0.001	0.003	<0.001	0.004
Middle spectrum energy (%)	17.60 (14.81, 24.67)	18.61 (12.06, 20.53)	11.90 (9.84, 17.85)	14.678	<0.001	—	<0.001	0.002
High spectrum energy (%)	14.30 (10.50, 24.07)	10.53 (6.96, 18.23)	6.40 (5.46, 11.74)	21.700	<0.001	—	<0.001	0.008

Note: —, $p > 0.05$.

The YBT showed that there were significant differences in the original and standard scores of A, PM, and comprehensive values among the three visual states ($p < .001$) (Table 2). The scores from highest to lowest was as follows: corrected vision, myopia, and eyes-closed ($p < .001$) (Table 2). The original and standard scores of PL between myopia and corrected vision state had no differences ($p > .05$) (Table 2).

The knee, and ankle joints under three visual states had prominent differences in the A direction ($p < .001$) (Table 3). The maximum flexion angle from largest to smallest was as follows: corrected vision, myopia, and eyes-closed. The hip joint had no difference A direction ($p > .05$) (Table 3). The hip and knee joints in three visual states were significant differences in the PM direction ($p < .001$) (Table 3). The joints angles had no differences between myopia and corrected vision state in the PL ($p > .05$) (Table 3).

The iEMG and RMS had significant differences in the three different states ($p < .01$) (Tables 4, 5) in the all directions of YBT.

4 Discussion

The purpose of this research was to explore the effects of different visual states on postural stability. The results showed that when maintaining static postural stability, the regularity of postural control in the myopia state was more regular than corrected vision. Under different visual acuity, the dynamic postural stability from stable to unstable was corrected vision, myopia, and eyes-closed. The activation and the work of the TA and GA from largest to smallest was as follows: eyes-closed, myopia, and corrected vision. Compared with corrected vision, the dynamic postural stability of myopia was decreased and the activation of ankle muscles was increased.

Analysis of the linear index revealed that the static postural stability in the state of corrected vision was better than that of eyes-closed. Entropy analysis of the non-linear index sample showed that when maintaining static postural stability, the postural control regularity under different visual conditions was as follows: eyes-closed, myopia, and corrected vision. That was, the attention required in the eyes-closed state for maintaining postural stability was higher than the states of myopia and corrected vision. Although the linear analysis method can describe and analyze the development and changes in postural control (Yamagata et al., 2017), the non-linear time series analysis method can provide a richer supplementary description of the potential developmental changes in postural control. SampEn is one of the various types of entropy measures that is used to evaluate the

TABLE 2 Comparison of extension scores of the Y-balance test in different visual states (mean ± standard deviations).

Project		Eyes-closed	Myopia	Corrected vision	*F*-value	*p*-value	Post-hoc comparisons		
							Eyes-closed VS. myopia	Eyes-closed VS. corrected vision	Myopia VS. corrected vision
							p-value	*p*-value	*p*-value
Standard value (%)	A	55.72 ± 9.30	62.77 ± 8.04	67.81 ± 8.24	56.656	<0.001	<0.001	<0.001	<0.001
	PM	92.13 ± 16.09	101.51 ± 11.73	108.86 ± 13.01	38.751	<0.001	<0.001	<0.001	0.003
	PL	86.19 ± 17.30	101.50 ± 15.14	104.37 ± 13.30	39.368	<0.001	<0.001	<0.001	—
	Composite	78.01 ± 12.90	88.59 ± 10.51	93.68 ± 10.53	78.429	<0.001	<0.001	<0.001	<0.001
Original Value (cm)	A	50.73 ± 7.42	57.31 ± 7.03	61.48 ± 7.10	50.938	<0.001	<0.001	<0.001	<0.001
	PM	84.13 ± 12.15	91.75 ± 10.29	98.07 ± 9.53	32.125	<0.001	0.003	<0.001	0.003
	PL	78.70 ± 15.32	91.56 ± 11.96	94.33 ± 9.41	37.972	<0.001	<0.001	<0.001	—
	Composite	213.57 ± 31.21	240.61 ± 25.93	253.89 ± 22.74	71.325	<0.001	<0.001	<0.001	<0.001

Note: —, $p > 0.05$.

variation in postural swing with time, behavior regularity, and predictability. It is generally believed that the lower the SampEn, the more regular the COP signal regularity. The lower the automation of postural control, the higher the requirement for attention (Roerdink et al., 2011; Kędziorek and Błażkiewicz., 2020). Therefore, when visual information is impaired, automation of body postural control and adjustment is reduced. To maintain postural stability, myopia required more attention than corrected vision. In a study of postural control in children and adolescents of different ages, Kiefer et al. (2021) found that there was no difference in the length of the COP trajectory in the young group with eyes open, but the regularity of postural control in the eyes-closed state was higher than that in eyes-opened states. However, studies on the effects of vision on static balance had revealed that the static balance of college students in normal vision states was better than that of those eyes-closed (Bednarczuk et al., 2021). In this study, COP data were only different in corrected vision and eyes-closed states, but there was no difference between COP data of myopia and those of other states. This may be because the participants recruited in this study were young male undergraduates, and the task of standing on two feet under different visual states was relatively simple and not challenging for them (Sarto et al., 2022). In a study on the effect of corrected vision on postural stability, Bae et al. (2020) found that the static postural stability of uncorrected visual acuity (without glasses) was significantly lower than that of the corrected state. People with myopia were more unstable than those with emmetropia in terms of visual-evoked postural response (Sayah et al., 2016). When myopia progresses to a certain extent, it causes blurring of visual information and decreases contrast sensitivity and stereoscopic perception (Ikuno, 2017), which leads to a decline in postural stability. Myopia and eyes-closed states also led to a significant increase in the middle and high-frequency spectral energy. Bae et al. (2020) found that myopia increased the middle- and high-frequency spectral energy of somatosensory compared with corrected vision, which also proved that myopia had a negative effect on static postural stability.

In this study, we observed that the original and standard values of the dynamic postural stability of the lower extremity were significantly different among the three visual states. The dynamic stability in the corrected vision state was higher than the myopia and eyes-closed states, and myopia state higher than eyes-closed. However, the stability of PL myopia was similar to that of corrected vision, which was higher than that of the eyes-closed state. In the YBT, the lower limb joint movement strategies were different according to different directions and vision. Muscle activation was also different, and the TA and GA were more obviously affected by vision in the three directions. The activation and work of the TA and

TABLE 3 The Y-balance test comparison of maximum flexion angle of lower limb joint in three visual states (mean ± standard deviations).

Project		Eyes-closed	Myopia	Corrected vision	*F*-value	*p*-value	Post-hoc comparisons		
							Eyes-closed VS. myopia	Eyes-closed VS. corrected vision	Myopia VS. corrected vision
							p-value	*p*-value	*p*-value
A	ankle	26.95 ± 8.91	31.56 ± 10.30	35.23 ± 9.39	17.408	<0.001	0.028	<0.001	0.013
	knee	45.24 ± 18.90	56.36 ± 21.89	64.85 ± 22.00	28.545	<0.001	0.001	<0.001	0.001
	hip	26.14 ± 16.00	25.61 ± 17.47	27.76 ± 21.69	0.409	0.584	—	—	—
PM	ankle	22.41 ± 11.82	24.48 ± 11.05	26.42 ± 10.94	10.933	<0.001	0.021	0.002	—
	knee	43.56 ± 20.05	52.11 ± 18.34	63.56 ± 20.42	34.913	<0.001	0.002	<0.001	<0.001
	hip	58.13 ± 19.34	69.95 ± 17.04	79.38 ± 18.54	32.627	<0.001	<0.001	<0.001	0.002
PL	ankle	20.25 ± 9.79	25.27 ± 12.18	28.01 ± 14.60	27.900	<0.001	0.003	<0.001	—
	knee	28.19 ± 16.88	40.50 ± 19.30	42.04 ± 21.68	15.268	<0.001	<0.001	<0.001	—
	hip	49.22 ± 15.08	65.44 ± 19.28	67.70 ± 18.55	24.490	<0.001	<0.001	<0.001	—

Note: —, $p > 0.05$.

TABLE 4 The Y-balance test RMS comparison of lower limb muscles in three visual states (Median values (25 percent of quartiles, 75 percent of quartiles)).

Project		Eyes-closed	Myopia	Corrected vision	χ^2-value	*p*-value	Post-hoc comparisons		
							Eyes-closed VS. myopia	Eyes-closed VS. corrected vision	Myopia VS. corrected vision
							p-value	*p*-value	*p*-value
A	TA	1.11 (0.65, 2.07)	0.78 (0.41, 0.53)	0.70 (0.37, 1.23)	30.900	<0.001	<0.001	<0.001	—
	GM	3.11 (1.48, 4.80)	2.60 (1.13, 3.70)	2.58 (1.08, 3.37)	16.300	<0.001	0.022	<0.001	—
PM	TA	0.83 (0.75, 2.24)	0.78 (0.50, 1.45)	0.60 (0.43, 1.31)	22.900	<.001	0.002	<0.001	—
	GM	2.08 (0.92, 3.53)	1.85 (0.78, 3.01)	1.55 (0.59, 2.35)	30.700	<0.001	0.008	<0.001	0.034
PL	TA	0.94 (0.72, 1.98)	0.72 (0.50, 1.69)	0.68 (0.40, 1.55)	30.700	<0.001	0.008	<0.001	0.034
	GM	2.16 (0.94, 3.19)	1.32 (0.65, 2.50)	1.08 (0.55, 2.10)	36.400	<0.001	0.002	<0.001	0.034

Note: —, $p > 0.05$.

TABLE 5 The Y-balance test iEMG comparison of lower limb muscles in three visual states (Median values (25 percent of quartiles, 75 percent of quartiles)).

Project		Eyes-closed	Myopia	Corrected vision	χ^2-value	*p*-value	Post-hoc comparisons		
							Eyes-closed VS. myopia	Eyes-closed VS. corrected vision	Myopia VS. corrected vision
							p-value	*p*-value	*p*-value
A	TA	58.09 (38.48, 99.73)	42.02 (27.62, 84.69)	37.99 (21.66, 71.36)	30.700	<0.001	0.008	<0.001	0.034
	GM	28.41 (19.38, 46.05)	25.26 (16.05, 42.81)	17.65 (11.82, 41.51)	19.900	<0.001	0.022	<0.001	—
PM	TA	54.13 (35.83, 71.08)	44.82 (27.78, 55.60)	39.21 (25.54, 50.69)	22.800	<0.001	0.013	<0.001	—
	GM	26.00 (17.10, 43.64)	19.36 (15.39, 33.23)	15.87 (11.66, 35.39)	28.900	<0.001	0.022	<0.001	0.022
PL	TA	65.90 (38.89, 104.33)	48.12 (33.74, 85.89)	37.64 (30.66, 64.98)	25.900	<0.001	0.008	<0.001	—
	GM	21.83 (14.69, 43.33)	18.61 (11.51, 40.38)	15.32 (10.74, 41.21)	12.700	0.002	0.005	0.008	—

Note: —, $p > 0.05$.

GA from largest to smallest was as follows: eyes-closed, myopia, and corrected vision states. Vision is the main sensory modality required by the body to maintain postural stability (Grace et al., 2012). When the body was in the states of eyes-closed or myopia, the immediate disappearance or inaccuracy of visual information altered sensory input and decreased postural stability. Vision had a greater influence on dynamic stability in the A and PM directions of the YBT, in which postural stability in myopia was worse than that in the corrected vision state but better than that in the eyes-closed state. It was further explained that blurred or impaired vision would have a negative effect on postural stability. However, in the PL direction, the postural stability in the states of myopia and corrected vision were similar but higher than that of the eyes-closed states. This may be because the PL structure of the YBT was not similar to the A direction, where the stretching baffle could be seen directly. The head and neck were generally in a neutral position during PL stretching, regardless of whether the vision was corrected, and there was a deviation between the afferent and feedback of PL spatial positioning and movement accuracy.

In the YBT, vision had a great influence on the range of motion of the ankle, knee, and hip joints of the lower extremity, in which there were significant differences in the maximum flexion angles of the ankle and knee joints under the three types of visual acuity, and the angles in the corrected vision state were larger than the myopia and eyes-closed states, and the myopia state larger than eyes-closed. Differences were observed between the PM and PL components in the hip joints. In adults, the extension scores of the YBT were moderately positively correlated with the ankle dorsiflexion angle (Kang et al., 2015a; Kang et al., 2015b), and the hip joint was mainly reflected in the PM and PL components (Kang et al., 2015a; Kang et al., 2015b), which also showed a consistent trend in this study.

The visual system also plays an important role in postural control. Schmidt et al. (2021) compared the effects of eye-opening, eyes-closed, wearing sunglasses, and a dark environment on postural stability, and found that dynamic stability was worse when the eyes were closed, and the activation of the TA and GA was more obvious. Gao et al. (2020) also found that after visual deprivation, the work of TA and GM was higher than those before visual deprivation. Although the test methods were different in the YBT, especially in the A and PM directions, it was also found that the TA and GA were more obviously affected by visual acuity. The activation and work of the TA and GA from the largest to the smallest was as follows: eyes-closed, myopia, and corrected vision states. The neuromuscular control system includes four aspects: muscle strength, reaction time, proprioception, and postural control. When postural stability was disturbed, the neuromuscular control system enhanced postural control ability by regulating muscle activation and coordination (Pan and Jin., 2012). Combined with the range of motion of the ankle joint in the three visual states, it was found that when visual information was limited, the neuromuscular control system increased the activation and work of the TA and GA, and improved proprioception of

the muscles around the ankle to meet the need for dynamic stability. In a study of the relationship between adolescent patients with myopia and abnormal body posture, Wang and Feng (2022) found that abnormal body posture increased the risk of myopia. Frequent reminders by teachers and increased physical activity could reduce cases of myopia caused by abnormal body posture. In this study, compared with corrected vision, myopia caused a decrease in dynamic postural stability and increased the activation of lower limb muscles, while the risk of falls was more related to dynamic stability (Ray et al., 2008; Jeter et al., 2015). Myopia may increase the risk of falls; therefore, it is particularly important to correct body posture, enhance physical exercise, and promptly prevent and control the degree of myopia.

This study mainly discussed the characteristics of postural stability in moderate myopic undergraduates in the state of myopia immediately after taking off their glasses. In the future, research should focus on a comparative analysis of the postural stability of myopia immediately after taking off the glasses and after adaptation (after a long period of adaptation). It should also explore the muscles around the knee and hip.

5 Conclusion

Moderate myopia has a negative effect on human postural stability. Compared with the corrected vision state, the stability of static posture in the state of myopia was decreased, and the postural regularity was more regular. Dynamic postural stability in the myopia state was also worse than that in the corrected vision state, and also increased activation and the work of muscles around the ankles.

Data availability statement

The original contributions presented in the study are included in the article/supplementary material, further inquiries can be directed to the corresponding author.

Ethics statement

The studies involving human participants were reviewed and approved by the ethical review board of the scientific research project of the Medical Ethics Committee of Binzhou Medical University. The patients/participants provided their written informed consent to participate in this study.

Author contributions

ZH and XX conceived the presented idea, developed the framework, and wrote the manuscript. ZH processed the data. XX provided critical feedback and contributed to the final version. All authors were involved in the final direction of the paper and contributed to the final version of the manuscript. All authors have read and agreed to the published version of the manuscript.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Effect of kinesio taping in combination with vibration treatment on college students' delayed-onset muscle soreness

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Objective: Kinesio taping (KT) and vibration treatment (VT) can alleviate delayed-onset muscle soreness (DOMS) to some extent. However, the literature reports on the difference between the two treatments, and whether a joint intervention (JI) works better than single treatments remains unknown. This study compares the effects of KT, VT and JI on DOMS in college students.

Methods: A total of 88 college students were randomly divided into the KT (KTG, n = 21), VT (VTG, n = 22), JI (JIG, n = 23) and control (CG, n = 22) groups. All subjects underwent DOMS moulding. The baseline; immediate and 24, 48 and 72 h visual analogue scale (VAS) scores and knee extensor maximum isometric voluntary contraction (MIVC) were determined.

Results: The intergroup comparison showed the following results. 1) The VAS scores of the four groups peaked at 24 h and gradually decreased. The immediate, 24 h and 48 h VAS scores followed the order: JIG > KTG and VTG > CG. The 72 h VAS score followed the order: KTG < VTG < JIG < CG. 2) The knee extension MIVC in the four groups was lowest at 24 h and then gradually increased. JIG had larger immediate MIVC than CG. KTG, VTG and JIG had larger 24 h MIVC than CG. JIG had larger 48 h MIVC than KTG and CG.

Conclusion: KT and VT can reduce muscle pain and strength loss caused by DOMS to varying degrees. VT is better than KT in improving pain. The combined intervention worked better than single interventions.

KEYWORDS

delayed-onset muscle soreness, kinesio taping, vibration treatment, pain, muscular strength

1 Introduction

Delayed-onset muscle soreness (DOMS) occurs from 8 h to 24 h after unaccustomed or intense centrifugal exercise and can peak and gradually relieve within 48–72 h, requiring approximately 1 week to restore (Afonso et al., 2021; Fleckenstein et al., 2021; Sonkodi et al., 2021; Torre et al., 2021). DOMS comes with a series of muscle structural, histological and biochemical changes, thus affecting physical fitness and exercise training effects. The increased risk of exercise injury (Akinci et al., 2020; Pupo et al., 2021) and reduced adverse effects caused by DOMS deserve concern.

In 1973, Dr. Kenso Kase invented the kinesio taping (KT) technology (Tran et al., 2021; Turgut et al., 2021). Previous studies confirmed the positive effect of KT on DOMS. KT intervention before DOMS moulding reduces bicep soreness in young men and accelerates the maximum isometric voluntary contraction (MIVC) recovery in muscles (Lee et al., 2015). KT intervention lowers the cold and thermal pain thresholds and the visual analogue scale (VAS)

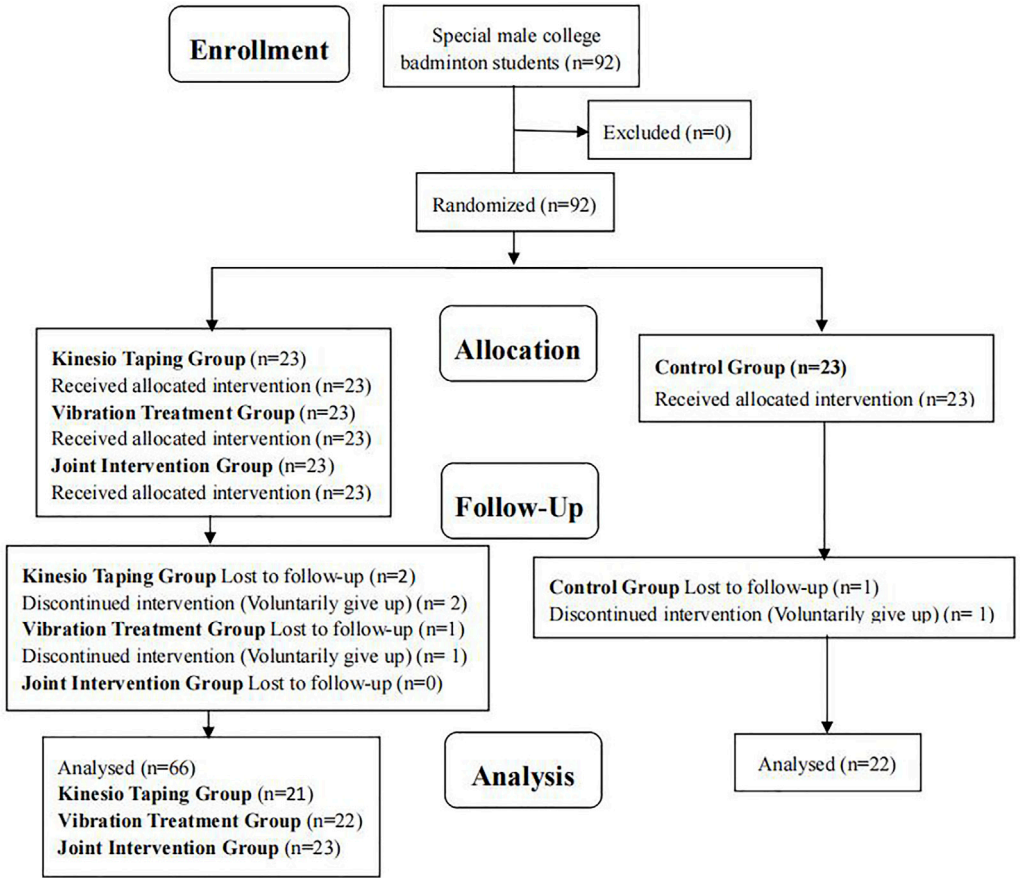

FIGURE 1
Participant selection flow diagram.

score for young male biceps brachii (DOMS) for 24 h (Bae et al., 2014). KT intervention also reduces centrifugal pain in the rectus femoris and hamstring (DOMS) for 24 and 168 h and improves the performance of hamstrings (Haksever et al., 2016). Furthermore, a meta-analysis suggested that appropriate vibration treatment (VT) can relieve DOMS symptoms (Lu et al., 2019). VT significantly reduces the VAS score and MIVC in ordinary young women (elbow DOMS, 50 Hz, 5 min) (Imtiyaz et al., 2014). The joint activity of ordinary male college students is increased (elbow DOMS, 20 Hz, 30 min), and the immediate VAS score (Lau and Nosaka, 2011) is reduced. MIVC for ordinary young males/females is increased (radial wrist extensor DOMS, 20 Hz, 2 min) (Koh et al., 2013). The VAS score of ordinary male college students is reduced (DOMS, 20 Hz–45 Hz, 10 min) (Wheeler and Jacobson, 2013). The above two methods can alleviate DOMS to some extent. KT can lift the skin of the ligation site, increase the tissue space and accelerate the blood and lymphatic circulation and inflammatory response (Tran et al., 2021). VT increases muscle discharge, local muscle temperature and blood flow to the skin; accelerates blood and lymphatic circulation and alleviates inflammatory response and pain (Lau and Nosaka, 2011; Koh et al., 2013; Wheeler and Jacobson, 2013; Imtiyaz et al., 2014; Lu et al., 2019). However, the literature only reports on the difference between the two treatments, and whether a joint intervention works better than single treatments remains unknown. Compared with single treatment (KT or VT), joint intervention may be a better way to delay DOMS.

This study explores the effect of KT with VT on DOMS symptoms in college students and provides the basis for the combined KT and VT intervention to reduce exercise-induced DOMS. The study hypothesises that the combination of KT and VT is more effective in relieving DOMS symptoms than a single treatments.

2 Materials and methods

2.1 Participants

This study was approved by our school's ethics committee (2021.011). Special male college badminton students were recruited during the holidays. The inclusion criteria were as follows: students (1) between 18 and 20 years old, (2) weighing 60 kg–70 kg, (3) exercising irregularly for the last 2 weeks and (4) who complied with the Declaration of Helsinki and provided informed consent. The exclusion criteria were as follows: students with (1) motor injury to the lower limbs, (2) heart pacemaker installation, (3) cardiovascular disease and (4) epilepsy or other central nervous system diseases.

This study based the sample size on previous research results regarding the previous study on Kinesio Taping and Vibration

TABLE 1 Participants' basic information.

Group	Age (years)	Height (cm)	Weight (kg)	BMI (kg/m^2)	VAS score	MIVC (Nm)
KTG (n = 21)	19.4 ± 0.2	176.1 ± 4.7	66.5 ± 4.1	21.4 ± 4.4	0.13 ± 0.03	191.2 ± 25.7
VTG (n = 22)	19.5 ± 0.2	176.0 ± 5.3	66.7 ± 3.8	21.5 ± 7.2	0.12 ± 0.08	193.4 ± 32.6
JIG (n = 23)	19.6 ± 0.2	176.3 ± 5.0	67.1 ± 3.9	21.6 ± 4.5	0.14 ± 0.08	188.2 ± 41.4
CG (n = 22)	19.5 ± 0.2	175.9 ± 4.6	66.3 ± 3.6	21.4 ± 4.1	0.15 ± 0.06	190.1 ± 23.5
F	0.332	0.215	0.423	0.341	0.292	0.354
P	0.769	0.823	0.661	0.711	0.746	0.692

Notes: KTG: kinesio taping group; VTG: vibration treatment group; JIG: joint intervention group; CG: Control group. VAS: Visual analog scale. MIVC: Maximum Isometric Voluntary Contraction.

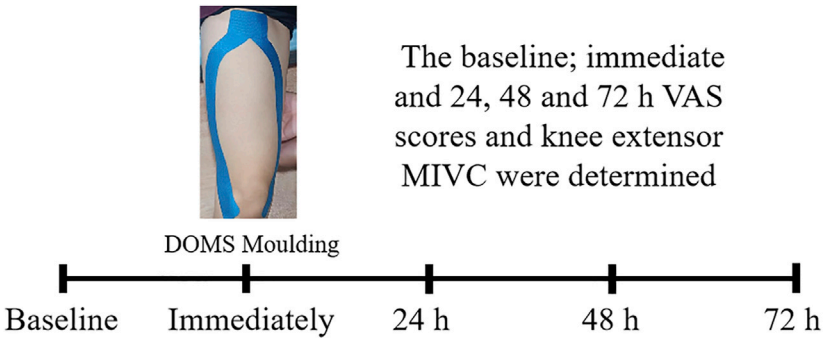

FIGURE 2
Intervention process.

Treatment in College Students'Delayed-Onset Muscle Soreness (Imtiyaz et al., 2014; Haksever et al., 2016), a 4 (groups) × 5 (measurement times) experimental design and a 10% sample wastage rate. G-power program was applied to calculate the effect size and total sample size. With an effect size of 0.3, α level of 0. 05 and power of 0.8, a sample size of at least 84 samples was needed. Ninety-two subjects were recruited. Four cases did not continue because of personal reasons. A total of 88 subjects completed the whole experiment (Figure 1). The random-number distribution method was used to divide subjects into four groups, i.e., KT (KTG), VT (VTG), joint intervention (JIG) and control (CG) groups. No significant difference in age, height and body weight was observed amongst groups (p > 0.05, Table 1). For VTG, VT was given after DOMS moulding. For KTG, Y-shaped binding was conducted for 15 min before DOMS moulding (Kirmizigil et al., 2019). After moulding, when VTG performed vibration intervention, KTG sat and rested. JIG, KTG and VTG interventions were simultaneous. For CG, no intervention happened except DOMS moulding.

2.2 DOMS moulding

In accordance with the previous method of DOMS knee moulding in the sports population (Pumpa et al., 2011; Song and Yang, 2022), all subjects underwent a maximum oxygen intake test 1 week before the moulding process (German Cortex cardiopulmonary function tester model: Metalyzer 3B). Moulding happened after a week, and subjects ran downhill on the treadmill (American ICON, model: 59916/350i; −10° grade, five groups of 8 min, with 2 min flat walking between groups). Each subject's exercise intensity was matched on the basis of the maximum oxygen intake, ensuring that all subjects maintained similar intensity while completing the downhill run. All participants were monitored to maintain their intensity at 80% of their maximum heart rate (Model No. M430; Polar, Finland) during downhill running. The treadmill speed was appropriately adjusted if the heart rate was excessively low or high. All participants were followed up within 72 h of DOMS moulding to ensure no other training or treatment (Figure 2).

2.3 Kinesio taping

The "Y" ligation of the bilateral knees of KTG and JIG was conducted by a professional physical therapist 15 min before DOMS moulding (Kirmizigil et al., 2019) (KT was provided by the LP Support company, specification: 5 cm × 5 m, color: blue), stretched to 125% of the original length (Boobphachart et al., 2017). (Figure 2). To prevent KT shedding throughout DOMS moulding and within 72 h, we checked immediately and at 24, 48 and 72 h with the same ligation.

2.4 Vibration treatment

Under the guidance of the experimenter, the American Power-Plate® pro5™ (71-PR5-3100, Frequencies 25–50 Hz) was used for

TABLE 2 VAS scores test for knee extension at different times.

Group	Baseline	Immediately	24 h	48 h	72 h
KTG (n = 21)	0.13 ± 0.03	1.86 ± 0.63^{ac}	2.30 ± 1.01^{ac}	1.28 ± 0.43^{aC}	0.60 ± 0.13^{BC}
VTG (n = 22)	0.12 ± 0.08	1.81 ± 0.70^{ac}	2.25 ± 0.89^{ac}	1.15 ± 0.31^{aC}	0.35 ± 0.11^{Ac}
JIG (n = 23)	0.14 ± 0.08	1.60 ± 0.56^{A}	1.72 ± 0.70^{A}	0.72 ± 0.29^{A}	0.13 ± 0.09^{A}
CG (n = 22)	0.15 ± 0.06	2.30 ± 0.61	2.62 ± 1.20	1.76 ± 1.00	0.72 ± 0.26
F	0.867	6.117	5.785	11.233	26.559
P	0.928	<0.01	0.003	<0.01	<0.01

Notes: VAS: visual analog scale; KTG: kinesio taping group; VTG: vibration treatment group; JIG: joint intervention group; CG: control group. KTG, VTG, and JIG, were compared at the same time points as CG: ^{a} $p < 0.05$; $^{A}p < 0.01$; KTG, was compared at the same time points as VTG; $^{B}p < 0.01$. KTG, and VTG, were compared at the same time points as JIG: ^{c} $p < 0.05$;.^{C} $p < 0.01$.

10 min on VTG and JIG (squatting and launching on the vibrational platform), referring to the vibration frequency interval and amplitude where VT had a positive effect on DOMS (Lau and Nosaka, 2011; Koh et al., 2013; Wheeler and Jacobson, 2013; Imtiyaz et al., 2014; Lu et al., 2019). This study's vibration frequency and amplitude were 45 Hz and 3 mm (G-Factor 7 45 Hz/High Amplitude), respectively. Four groups of subjects underwent VAS score and knee extension maximum isometric muscle strength tests at DOMS baseline or before moulding and after moulding, 24, 48 and 72 h.

2.5 VAS score

The VAS score was used to evaluate the pain of the DOMS (quadriceps femoris) to allow the subject to draw a vertical line on a 10 cm line (0 cm, painless; 10 cm, most painful; 0–10 and left and right knee averaging) (Cheng and Jiang, 2020; Arieh et al., 2022).

2.6 MIVC test for knee extension

The MIVC test was performed on left and right knee joints. The subject took a seated position, and the trunk and hip joints were fixed using a wide band. During the test, the subject underwent MIVC at knee flexion 30°, and the subject makes full effort (using the maximum force as far as possible) to extend the knee joint. The test time is 15 s, at this time, the maximum peak torque recorded by the isokinetic instrument is MIVC. The left and right knees of each subject were tested. The analysis index was peak torque (Cheng and Jiang, 2020; Cheng et al., 2021), and the left and right knee data of all subjects were averaged.

2.7 Statistical analysis

SPSS 19.0 software, mean ± standard deviation for four groups. A two-way analysis of variance with mixed design was used to test whether groups, the main effect of time, and group (4) and time (5) interacted. If group and time interacted, the one-way ANOVA with repeated measures would vary at different time points. If time had the main effect, different time points were compared. If groups had a main effect, differences between groups were compared. The Bonferroni for posthoc comparisons was adjusted to ensure that the overall type I rate per ANOVA was not greater than 0.05. The significance level was 0.05.

3 Results

The results of five time points for the four subjects are shown in Table 2; Table 3. First, the two-way ANOVA VAS score and knee extension MIVC group × time interacted ($p < 0.01$), showing differences at different time points within the one-way ANOVA group. The comparison of baseline data between groups showed that VAS scores ($F_{(3,\ 84)} = 0.292$, $p = 0.746$) and MIVC ($F_{(3,\ 84)} = 0.354$, $p = 0.692$) were not significantly different ($p > 0.05$, Table 1).

VAS score: Table 2 shows that the VAS score of the four groups peaked and gradually decreased at 24 h after DOMS moulding. The comparison between groups showed the following results. KTG and VTG had lower immediate, 24 h and 48 h VAS scores than CG ($p < 0.05$). JIG had lower immediate, 24 h and 48 h VAS scores than CG ($p < 0.01$). KTG and VTG had higher immediate, 24 h and 48 h VAS scores than JIG ($p < 0.05$). VTG and JIG had lower 72 h VAS score than CG ($p < 0.01$). KTG had higher 72 h VAS score than VTG ($p < 0.01$). KTG ($p < 0.01$) and VTG ($p < 0.05$) had higher 72 h VAS score than JIG.

MIVC: Table 3 shows that the MIVC of the four groups was lowest at 24 h after DOMS moulding. The comparison between groups showed the following results. JIG had larger immediate MIVC than CG ($p < 0.05$). KTG, VTG and JIG had larger 24 h MIVC than CG ($p < 0.05$). JIG had larger 48 h MIVC than KTG and CG ($p < 0.05$).

4 Discussion

This study seeks to explore pain and MIVC changes caused by KT and VT on special college badminton students' DOMS and provides the basis for reducing exercise-induced DOMS. We have tested the hypothesis that the combination of KT and VT is more effective in alleviating DOMS symptoms than single treatment.

4.1 Kinesio taping

This study found that KT significantly reduces immediate, 24 h and 48 h VAS scores. This finding confirms the results of previous similar studies on KT, which show that biceps brachii and DOMS moulding reduce immediate, 24 h, 48 h (Bae et al., 2014) and 72 h (Lee et al., 2015) VAS scores. KT is equally effective when subjects are female, and the DOMS site is the thigh, thus reducing the VAS score

TABLE 3 MIVC Test for Knee Extension at different times. (Nm).

Group	Baseline	Immediately	24 h	48 h	72 h
KTG (n = 21)	191.2 ± 25.7	177.7 ± 28.9	169.2 ± 27.8[a]	176.2 ± 26.4[c]	187.6 ± 36.2
VTG (n = 22)	193.4 ± 32.6	173.6 ± 25.4	172.6 ± 32.5[a]	178.1 ± 34.0	190.5 ± 28.7
JIG (n = 23)	188.2 ± 41.4	184.2 ± 23.0[a]	180.9 ± 28.7[a]	190.0 ± 27.5[a]	190.2 ± 38.6
CG (n = 22)	190.1 ± 23.5	168.2 ± 27.1	160.5 ± 21.2	177.3 ± 26.8	186.3 ± 21.0
F	0.725	2.868	2.919	2.665	0.362
P	0.830	0.020	0.013	0.032	0.701

Notes: MIVC: maximum isometric voluntary contraction; KTG: kinesio taping group; VTG: vibration treatment group; JIG: joint intervention group; CG: Control group. VAS: visual analog scale. KTG, VTG, and JIG, were compared at the same time points as CG: [a] $p < 0.05$; KTG, and VTG, were compared at the same time points as JIG: [c] $p < 0.05$.

for the rectus femoris and hamstrings at 48 and 168 h in young women (Haksever et al., 2016).

Why does KT reduce muscle pain caused by DOMS? The vessel diameter of the DOMS site decreases, thus impairing blood flow reaction and causing hyperalgesia (Ko and Clarkson, 2020). Tissue swelling at the DOMS damage site affects blood and lymph circulation (Fang and Nasir, 2021). KT can lift the skin of the ligation site, increase the tissue space and accelerate the blood and lymphatic circulation and the inflammatory response. This lifting can help relieve pressure on nociceptors directly under the skin and remove the accumulated metabolites (Tran et al., 2021), thus reducing pain and feeling. Alternatively, KT provides sustained tactile and proprioceptive inputs to the ligation site, thereby inhibiting the input of nociception (Saki et al., 2022).

This study found that KT significantly enhances the MIVC of the knee extensor for 24 h. The meta-analysis (including four randomised controlled trials) confirmed our conclusion that KT significantly increases the knee extensor muscle (DOMS) MIVC (Tran et al., 2021). KT increases skin input then enhances the function of neuromuscles, promotes muscle activity at the ligation site, improves muscle structure and contributes to a slight increase in muscle force (Tran et al., 2021). Furthermore, the subject reduces the muscle strength test caused by pain (does not exert more force).

4.2 Vibration treatment

This study found that VT significantly reduces immediate, 24 h, 48 h and 72 h VAS scores. Previous studies showed that VT reduces muscle pain in the knee joint immediately after DOMS. VT significantly reduces the immediate, 24 h and 48 h VAS scores (20 Hz–45 Hz, 10 min) of ordinary college male students (Wheeler and Jacobson, 2013) and significantly reduces the VAS scores (30 Hz, 10 min) of men at 96 and 120 h (Broadbent et al., 2010). In the present study, VT at 45 Hz for 10 min is adopted for DOMS sites. The vibration frequency and time of the intervention vary, and the VAS score is significantly reduced at different measurement times. These findings are similar to those of previous studies (Broadbent et al., 2010; Wheeler and Jacobson, 2013), showing that VT can relieve muscle pain after DOMS moulding.

Several current views explained the pain reduction caused by VT. VT stimulates myospindle and motor neurons, causing strong muscle contraction to produce a perceptual response and relieve pain. VT activates large-muscle diameter fibres, inhibits small-diameter fibres and reduces pain. VT also accelerates the stimulation of inhibitory interneurons in the middle nerve endings of the spinal nerve and reduces the transmission of perceptual information from the spinal cord to the brain (feedback of proprioceptors for nociception inhibition) to reduce pain. VT increases muscle discharge, local muscle temperature and blood flow to the skin; accelerates blood and lymphatic circulation and alleviates the inflammatory response and pain (Lau and Nosaka, 2011; Koh et al., 2013; Wheeler and Jacobson, 2013; Imtiyaz et al., 2014; Lu et al., 2019).

VTG has significantly higher MIVC than CG (24 h). In previous studies, VT (50 Hz, 5 min) on the elbow in ordinary young women shows that the MIVC increases at 72 h (Koh et al., 2013). VT (50 Hz, 30 min) on common male/female femoral quadriceps reveals a significant increase in MIVC for 24 h (Bakhtiary et al., 2007). The above study partly supports the present study's conclusions. The analysis may be related to the activation of increased motor units (DOMS sites) in increased muscle tension. Muscle spindle stimulation by vibration may increase the afferent activities of muscle spindles, which may increase background tension in the vibrated muscles (Bakhtiary et al., 2007; Broadbent et al., 2010; Koh et al., 2013). Furthermore, given the reduced muscle pain, the subject is believed to have reduced muscle strength caused by pain (afraid to force) during the isometric muscle force test.

4.3 Joint intervention

This study has compared KTG, VTG and JIG. The VAS score of JIG at 72 h decreases significantly compared with that of the control group. The MIVC of the extensor knee muscles is significantly increased at 24 and 48 h. The VAS scores of JIG immediate ~72 h are considerably smaller than those of KTG and VTG, indicating that the combined intervention can better relieve the muscle pain caused by DOMS than single treatments. The MIVC of JIG is significantly higher than that of KTG (48 h), indicating that the combined intervention significantly affects muscle force at 48 h than single treatments. No literature report has been made on combining conditions on muscle pain and strength at the DOMS site. The effects are also speculated to increase after the joint intervention of KT and VT, and we can provide evidence for similar future studies. This study has found that the VAS score (72 h) of VTG is less than that of KTG, indicating that VT relieves muscle pain caused by DOMS better than KT over time. Our results suggest that, in clinical practice, reasonable KT and VT

intervention may delay the symptoms of DOMS when the exercise population performs high-intensity exercise.

This study has some limitations. First is the small sample size, which only comprise special college badminton students. Second, no blood indicator is collected. Lastly, a placebo treatment is not designed. Further expansion of the sample is needed to explore how KT and VT improve DOMS symptoms.

5 Conclusion

KT and VT can significantly reduce muscle pain and strength loss caused by DOMS. VT works better than KT in improving pain. The combination treatment is better than single interventions.

Data availability statement

The raw data supporting the conclusion of this article will be made available by the authors, without undue reservation.

Ethics statement

This study was approved by the ethics committee of Chengdu Sport University (2021.011). Before the assessment, every participant received the same detailed information about the testing procedure. Every participant signed the informed consent. The patients/participants provided their written informed consent to participate in this study.

Author contributions

Author contributions NQ: designing this study and writing initial draft and revision. NQ: making figure and table. NQ: rechecking the manuscript and putting forward suggestions for amendment. NQ: revising language and content, supervision, project administration, and funding acquisition. NQ contributed to the article and approved the submitted version.

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Conflict of interest

The author declares that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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The effect of Bafa Wubu of Tai Chi on college students' anxiety and depression: A randomized, controlled pilot study

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Objective: This pilot study aimed to explore the mechanism of the effects of Bafa Wubu of Tai Chi (BWTC) on anxiety and depression in college students using resting-state functional magnetic resonance imaging (RS-fMRI).

Methods: Eighteen college students (5 males and 13 females) with anxiety and depression met the study criteria and were randomly divided into an experimental group (aged 24.20 ± 4.07 years) and a control group (aged 22.50 ± 5.95). The experimental group received an eight-week BWTC intervention five times/week for 60 min/session. The control group maintained normal daily life without any exercise intervention. These students were assessed using RS-fMRI scans, the self-rating anxiety scale (SAS), and the self-rating depression scale (SDS). Spearman correlation analysis was used, and statistical significance was defined as a two-sided *p*-value of <0.05.

Results: After the intervention, the SAS and SDS scores of the BWTC group significantly reduced ($p = 0.002$; $p = 0.001$). Compared with the control group, the fALFF values of the right middle frontal gyrus, orbital part (Frontal_Mid_Orb_R) ($p = 0.043$), right inferior occipital gyrus (Occipital_Inf_R) ($p = 0.003$), and right middle temporal gyrus of the temporal pole (Temporal_Pole_Mid_R) ($p = 0.003$) in the BWTC group increased significantly; the fALFF values of the left middle frontal gyrus (Frontal_Mid_L) ($p = 0.001$) and right supplementary motor area (Supp_Motor_Area_R) ($p = 0.010$) in BWTC group decreased significantly. The fALFF values of Frontal_Mid_Orb_R were significantly positively correlated with the SDS score ($r = 0.852$, $p = 0.015$) and the fALFF values of Frontal_Mid_L were significantly negatively correlated with the SAS score ($r = -0.797$, $p = 0.032$).

Conclusion: In this pilot study with college students, BWTC alleviated anxiety and depression, potentially through modulating activity in the Frontal_Mid_L and Frontal_Mid_Orb_R, respectively.

KEYWORDS

Bafa Wubu of Tai Chi, anxiety, depression, college students', RS-fMRI

1 Introduction

According to the theory of emerging adulthood, college students belong to a unique developmental phase spanning from late adolescence to adulthood (aged 18–29 years) (Arnett, 2000; Jin et al., 2023). College students face a wide range of changes in cognitive function (O'Rourke et al., 2020; Taylor and Snyder, 2021), emotional regulation, and behavior, which are linked to a greater possibility of engaging in risky behaviors e.g., Internet (or smartphone) addiction (Lu et al., 2020), alcohol abuse (Nelson and Padilla-Walker, 2013), alcohol-related sexual behavior (Litt et al., 2020; Persike et al., 2020; Edgerton and Keough, 2021), and developing mental health issues e.g., distorted eating habits, depression, and anxiety (Chi et al., 2020; Lin et al., 2020; Berge et al., 2021; Zhang et al., 2022) due to poor adjustment during this transitional period. Notably, anxiety and depression among college students are the most reported negative emotions considered as risk factors that affect physical (Fuller-Rowell et al., 2021) and mental health (Guastaferro and Bray, 2020; Halliburton et al., 2021), social function (Reed-Fitzke et al., 2021), academic performance (Gillen-O'Neel et al., 2021; Haikalis et al., 2022, and later career success/employment (Masdonati et al., 2022). Previous studies have indicated that 20%–30% of college students report different levels of anxiety and depression (Islam et al., 2020). The two intertwined negative emotions do not simply place a significant financial burden on individuals and their families but also challenge the national healthcare system (Simon et al., 1995). Therefore, there is an urgent need to identify effective strategies to prevent and treat these mental health issues.

Accumulating evidence indicates that non-pharmacological intervention (e.g., cognitive behavioral therapy, mindfulness/mediation, and mindful exercise) has beneficial effects on relieving anxiety and depressive symptoms (Neufeld et al., 2020; Chi et al., 2021; Pardos-Gascón et al., 2021; Pérez-Aranda et al., 2021). Compared to Western psychotherapy, Eastern medicine, especially traditional Chinese medicine (TCM), has received great attention over the past two decades. Within the theory of Chinese medicine, traditional Chinese mind-body exercise (e.g., Tai Chi, Baduanjin, and Wuqingxi) characterized by slow eye-hand and eye-foot coordinative movement, rhythmic deep breathing, and meditative state (Yeung et al., 2018) has become more popular worldwide (Zou et al., 2017; Zou et al., 2018). Notably, these beneficial features of Tai Chi are suitable for vulnerable populations, such as college students suffering from anxiety and depressive symptoms in school settings (Chang et al., 2021). For example, a seminar meta-analysis by Zhang et al. (2019) has concluded that Tai Chi is a cost-effective non-pharmacological approach for relieving anxiety and depressive symptoms among non-clinical individuals across different age groups including college students.

Notably, the potential neurobiological mechanisms underlying the alleviative effects of Tai Chi on depression and anxiety are still not fully understood. With emerging imaging techniques in recent years, exercise scientists have developed new tools to study changes in brain activity, structure, and function due to Tai Chi practice-induced changes (Yue et al., 2020a; Yue et al., 2020b; Chen et al., 2020; Yue et al., 2020c; Yu et al., 2021). Of note, previous imaging studies rarely used resting-state functional magnetic resonance imaging (RS-fMRI), which, as an emerging data acquisition tool, can measure functional connections and local neural activity in the brain in response to changing emotional states (Sparacia et al., 2019). This study used RS-fMRI to explore the effects of Bafa Wubu of Tai Chi (BWTC) on anxiety and depression in college students, and determine whether BWTC can effectively alleviate these negative emotions. This pilot study hypothesized that BWTC could decrease the levels of anxiety and depression in college students and that the neurological mechanism could be examined by fMRI.

2 Materials and methods

2.1 Participants

A total of 33 college students who reported different levels of anxiety and depression were recruited from the Student Psychological Counseling and Service Center of Beijing Normal University. After evaluation according to the inclusion and exclusion criteria, only 18 college students (five males and 13 females) were retained in the sample. The participants reported an age range of 18–30 years old. They voluntarily agreed to participate in this study and signed informed consent forms. The inclusion criteria were as follows: 1) a self-rating anxiety scale (SAS) (Zung, 1971) score of no less than 50, and 2) a self-rating depression scale (SDS) (Zung et al., 1965) score of no less than 53. The exclusion criteria were as follows: 1) major physical illness or serious mental illness, 2) taking medication, 3) movement disorders and serious joint damage, and 4) presence of metal objects in the body, such as pacemakers.

2.2 Study design

This pilot study involved college students. The participants were randomly assigned into two groups (1:1 ratio): the BWTC and control groups. The intervention lasted for 8 weeks. The research process included recruitment, screening, randomization, intervention, and follow-up. This study was approved by the Experimental Ethics Committee of the Department of Psychology of Beijing Normal University and conducted in compliance with the Declaration of Helsinki.

2.3 SAS and SDS

SAS was used to test students' anxiety. SAS was tested for reliability with an intraclass correlation coefficient (ICC) of 0.908 (Zung, 1971). Participants were asked to rate 20 items on a scale of 1–4. The total score for each participant was then multiplied by 1.25. The anxiety score showed a positive correlation with anxiety tendency, and the anxiety of a participant was considered a critical value if his/her score reached 50 (Olatunji et al., 2006).

The SDS was used to test participants' depression. SDS was tested for reliability with an intraclass correlation coefficient (ICC) of 0.937 (Zung et al., 1965). The main statistical indices, scoring criteria, and statistical methods employed by this scale were the same as those used by the anxiety self-rating scale. The depression score showed a positive correlation with depression tendency, and the depression of a

participant was considered a critical value if his/her score reached 53 (Wittkamp et al., 2018).

2.4 Sample size calculation

SAS and SDS were set as coprimary outcomes and used for sample size calculation. The sample size was calculated based on the changes in the SAS and SDS between comparison groups with a significance level of 5% and a two-tailed critical region to detect an effect size of Cohen's d = 0.50, with 80% power using G*Power V.3.1.9.6 software. Post-intervention, the mean scores of SAS and SDS in the control and intervention groups were (43.9 ± 5.6, 32.3 ± 4.5) and (53.6 ± 8.7, 39.2 ± 9.3) (Li et al., 2019), respectively, according to the published literature. Because the sample size calculation of SAS was smaller than that of SDS, the sample size calculation of SDS was selected. This would require 12 participants, increased to 18 to account for the failure to follow-up approximately 30% of participants, with nine participants assigned to each group.

2.5 Randomization and blinding

Eighteen participants were randomized into groups using SPSS 21.0 software. The steps were as follows: 1) Encode 18 participants 1–18, 2) set the random number of seeds (0–2000000), 3) generate random numbers (ranged 0–1), 4) sort and group (using the visual discretization method, the number of segmentation points is 1, the width is 50%, and the cut point value is 0.318), and 5) the results of the random grouping were obtained. Blindfolding the participants was difficult because of the characteristics of the intervention measures. Hence, only indicator testers and data analysts were blindfolded.

2.6 Intervention plans

The college students in the intervention group received a 1-h BWTC exercise intervention five times a week for eight weeks. The intervention involved 10 min of warm-up activities, 40 min of practice, and 10 min of closing activities. The intervention emphasized not only the postures but also the combination of body, breath, and mind, given that BWTC requires one to "build his/her body," "convey his/her breaths," and "use his/her mind" (Lyu, 2018). Participants in the control group went on with their daily lives as usual and did not perform regular physical activities. None of the participants were taking psychotropic or other drugs with similar effects during the study.

2.7 Data acquisition

Data were collected at the Magnetic Resonance Laboratory of the Magnetic Resonance Imaging Research Center of Peking University. A Siemens 3tPrisma MRI scanner was used for the data collection. The MRI real radiology department was equipped with an eight-channel phased array head MRI system, which was also used for the acquisition. Functional imaging was performed using gradient echo planar imaging sequences. The scanning parameters obtained included repetition time (TR) = 2,000 ms, echo time (TE) = 30 ms, field of view (FOV) = 220 mm × 220 mm, flip angle (flip angle) = 90°C, matrix = 64 × 64, number of layers = 36, layer thickness = 4 mm, and scan time = 480 s. To mitigate noise from the scanner, each participant was given earplugs before the test. At the beginning of the test, the participants lay flat on the scanner with their heads fixed. They were required to keep their heads and bodies still and remain in an awake and relaxed state throughout the test.

2.8 Data processing

The data were analyzed and processed on the Matlab13b platform using the Data Processing Assistant for RS-fMRI software. The data were processed as follows: 1) data format conversion: the data format was converted from Digital Imaging and Communications in Medicine (DICOM) to the Neuroimaging Informatics Technology Initiative (NIfTI); 2) removal: the data for the first 10 time points were removed (the test involved 240 time points, and only 230 time points were used as experimental data); 3) slice timing: the middle of the total number of layers was set; 4) realign: according to the head movement correction curve, participants showing a U-turn movement in the x-, y-, and *z*-axes with translations greater than 1 mm and rotations greater than 1°C were excluded; 5) normalize: the inter-individual differences were reduced to an MNI ERI template through spatial normalization. The voxel resampling was 3 mm × 3 mm × 3 mm; 6) smooth: the Gaussian smoothing kernel with a full width at half maximum (FWHM) of 6 mm was used to smoothen the noise reduction; 7) detrend: the temperature increases due to the work of the machine or the participants were adapted, and the accumulated data collection time showed a linear trend; and 8) calculate the ratio of low-frequency amplitude fraction (fALFF): the root of the power spectrum of the 0.01 Hz–0.08 Hz signal was used to obtain the ALFF value, and the ratio of the amplitude sum in the low-frequency band to the amplitude sum of the whole frequency band was used to obtain the fALFF value.

2.9 Statistical analysis

The SPSS 21.0 statistical software was used for the data processing. Continuous variables are described as mean ± standard deviation (SD) for normal distributions or median for non-normal distributions, and categorical variables are described as frequency. The baseline data mainly describe the characteristics of the participants. Two-way repeated measures ANOVA was used to analyze the effects of anxiety and depression before and after the intervention in the two groups and the interaction between them. Statistical parametric mapping (SPM12) software was used for statistical analysis. The fALFF values of different brain regions were extracted and *post hoc* tests were performed. Age, sex, height, and weight were used as covariates, and the Bonferroni correction was used. Paired-sample *t*-tests were used for within-group comparisons, and independent-sample *t*-tests were used for between-group comparisons. Within-group comparisons were

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TABLE 1 Baseline sample characteristics.

Items	BWTC group (n = 9)	Control group (n = 9)	*p-value*
Age (years)	24.20 ± 4.07	22.50 ± 5.95	0.483
Sex (n,%)	Male 2 (22.2%)	Male 3 (33.3%)	0.736
	Female 7 (77.8%)	Female 6 (66.7%)	
Height (cm)	167.90 ± 5.06	165.63 ± 8.07	0.475
Weight (kg)	61.78 ± 11.78	57.50 ± 11.46	0.500
SAS (score)	56.94 ± 9.23	54.58 ± 9.70	0.604
SDS (score)	61.80 ± 10.53	57.50 ± 14.92	0.490

Note: SAS: self-rating anxiety scale; SDS: self-rating depression scale.

TABLE 2 Two-way repeated measures ANOVA of SAS and SDS.

Indicators	Group		Time		Group × time	
SAS	F	*p*	F	*p*	F	*p*
	0.159	0.704	26.975	0.002	1.660	0.245
SDS	F	*p*	F	*p*	F	*p*
	0.305	0.601	41.324	0.001	2.235	0.186

Note: SAS: self-rating anxiety scale; SDS: self-rating depression scale.

performed with Family Wise Error (FWE) correction, and the union of one-sample *t*-test results of the two groups were used as the range of brain functional map comparisons ($p < 0.05$, number of voxels > 20). Comparisons between groups were performed with the FWE correction, and brain regions with $p < 0.05$ and the number of voxels > 20 were defined as statistically different regions. We used xjview and BrainNet Viewer software to present the results. Spearman correlation analysis was used to analyze the correlation between changes in brain area parameters and SAS and SDS scores. Statistical significance was defined as a two-sided *p*-value of <0.05.

3 Results

3.1 Baseline sample characteristics

Before the intervention, no significant differences were observed between the two groups in terms of age ($p = 0.483$), sex ($p = 0.736$), height ($p = 0.475$), weight ($p = 0.500$), SAS score ($p = 0.604$), or SDS ($p = 0.490$) (Table 1).

3.2 Comparison of SAS and SDS scores in the two groups

Compared to the change in SAS, there was no interaction between group and time (F = 1.660, $p = 0.245$). The results showed that the change in SAS scores had a main effect of time (F = 26.975, $p = 0.002$). Compared with before intervention, SAS in the BWTC group was significantly reduced after intervention (56.94 ± 9.23 vs. 41.60 ± 8.85, $p < 0.01$), while there was no significant difference in the control group ($p > 0.05$). Compared to the change in SDS, there was no interaction between group and time (F = 2.235, $p = 0.186$). The results showed that the change in SDS had a main effect of time (F = 41.324, $p = 0.001$). Compared with before intervention, SDS in the BWTC group was significantly reduced after intervention (69.29 ± 12.16 vs. 43.39 ± 10.17, $p < 0.01$), while there was no significant difference in the control group ($p > 0.05$). The results are presented in Tables 2, 3.

3.3 Comparison of RS-fMRI between the two groups after the intervention

After the intervention, the interaction between the fALFF values in the right middle frontal gyrus, orbital part (Frontal_Mid_Orb_R), left middle frontal gyrus (Frontal_Mid_L), right superior frontal gyrus (Frontal_Sup_R), right inferior occipital

TABLE 3 Comparison of the SAS and SDS scores of the two groups (_x ± s).

Indicators	Time	N	BWTC group	Control group
SAS	Before intervention	9	56.94 ± 9.23	54.58 ± 9.70
	After intervention	9	41.60 ± 8.85**	47.52 ± 8.72
SDS	Before intervention	9	69.29 ± 12.16	57.50 ± 11.32
	After intervention	9	43.39 ± 10.17**	51.80 ± 12.16

Note: SAS: self-rating anxiety scale; SDS: self-rating depression scale. Compared with before intervention, $^{**}p < 0.01$.

TABLE 4 Comparison of the fALFF values between the two groups after the intervention.

	Brain areas	Number of voxels	MNI			F
			X	Y	Z	
1	Frontal_Mid_Orb_R	64	0	27	−12	50.2123
2	Frontal_Mid_L	40	−30	6	60	51.55
3	Frontal_Sup_R	76	24	63	0	150.71
4	Insula_R	76	36	15	12	73.68
5	Occipital_Inf_R	59	45	−78	−3	84.61
6	Occipital_Sup_R	31	21	−84	39	43.81
7	Supp_Motor_Area_R	20	9	21	51	39.23
8	Temporal_Inf_L	22	−39	−3	−30	66.21
9	Temporal_Pole_Mid_R	79	21	−3	−30	67.23

Note: Results of the independent sample *t*-test, $p < 0.05$, number of voxels > 20, Family wise error correction. X: left and right; Y: front and back; Z: up and down; and F: the t-value of the Peak point, the one with the most significant difference in this brain region. fALFF: low-frequency amplitude fraction. The MNI, coordinate values are based on the Montreal Institute standard template space. Frontal_Mid_Orb_R: right middle frontal gyrus, orbital part; Frontal_Mid_L: left middle frontal gyrus; Frontal_Sup_R: right superior frontal gyrus; Insula_R: right insula; Occipital_Inf_R: right inferior occipital gyrus; Occipital_Sup_R: right superior occipital gyrus; Supp_Motor_Area_R: right supplementary motor area; Temporal_Inf_L: left inferior temporal gyrus; Temporal_Pole_Mid_R: right middle temporal gyrus of the temporal pole.

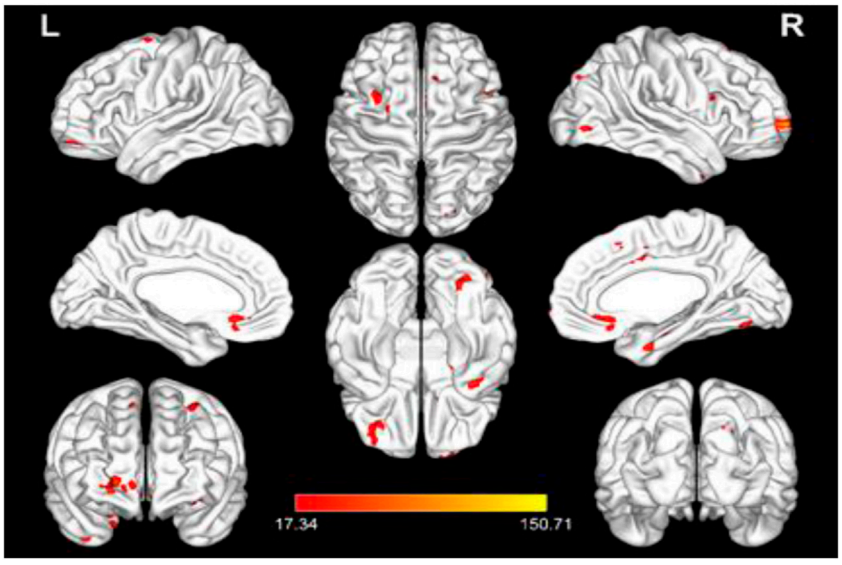

FIGURE 1
3D plot of the significant interaction of fALFF values. Note: The red bars represent those brain regions with significant interactions, and the statistical threshold was set to $p < 0.001$ and the number of voxels was > 20. L: left; R: right; and fALFF: low-frequency amplitude fraction.

gyrus (Occipital_Inf_R), right superior occipital gyrus (Occipital_Sup_R), right supplementary motor area (Supp_Motor_Area_R), left inferior temporal gyrus (Temporal_Inf_L), and right middle temporal gyrus of the temporal pole (Temporal_Pole_Mid_R) of the two groups changed significantly. The results are shown in Table 4; Figures 1, 2.

To further analyze the differences between the groups before and after the intervention in the same brain region, the fALFF values of different brain regions were extracted for *post hoc* analysis. After the intervention, compared with the control group, the fALFF values of the Frontal_Mid_Orb_R ($p = 0.043$), Occipital_Inf_R ($p = 0.003$), and Temporal_Pole_Mid_R ($p = 0.003$) in the BWTC group increased significantly, indicating that the local activity of the corresponding brain regions was enhanced, while the fALFF values of Frontal_Mid_L ($p = 0.001$) and Supp_Motor_Area_R ($p = 0.010$) in the BWTC group decreased significantly, indicating that the local activity of the corresponding brain regions was weakened. The results are shown in Figure 3.

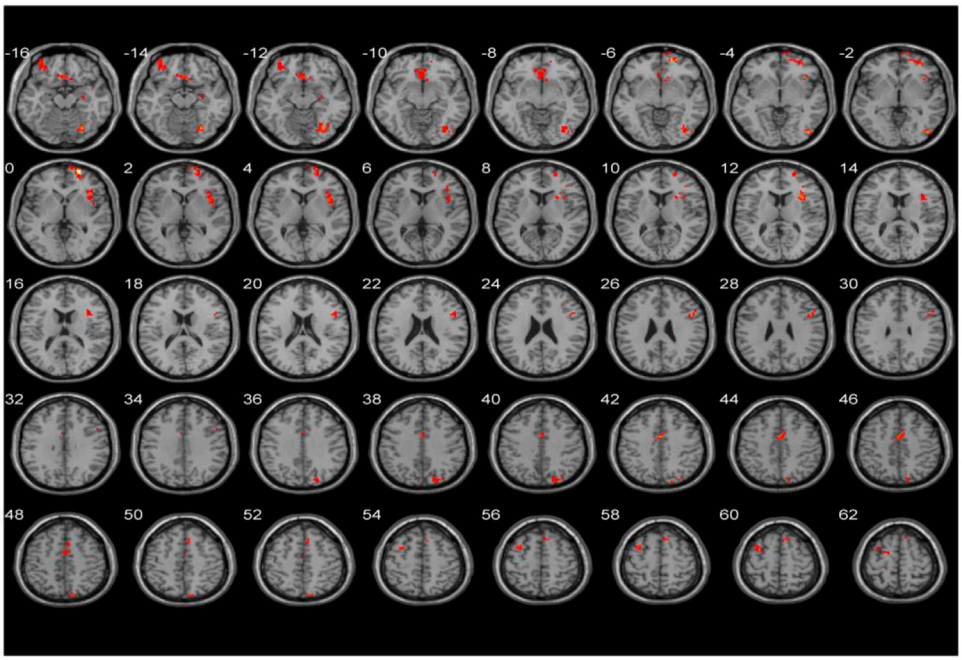

FIGURE 2
Significant plane diagram of the interaction effect of fALFF values. Note: The color bar represents the brain area with significant interaction, the statistical threshold was set to $p < 0.001$, and the number of voxels was > 20. fALFF: low-frequency amplitude fraction.

To further explore the potential correlation between anxiety and depression and local brain activity after the intervention, the correlation coefficients of fALFF values and SAS-SDS scores were calculated using Spearman correlation analysis. The fALFF values of Frontal_Mid_Orb_R were significantly positively correlated with the SDS score ($r = 0.852$, $p = 0.015$) and the fALFF values of Frontal_Mid_L were significantly negatively correlated with the SAS score ($r = -0.797$, $p = 0.032$) (Table 5).

4 Discussion

To the best of our knowledge, this study is the first to explore the mechanism underlying the effects of BWTC on anxiety and depression in college students using RS-fMRI. After eight weeks of BWTC, the participants in the BWTC group showed significant improvements in anxiety and depression, whereas those in the control group showed no significant changes. The RS-fMRI results showed that, compared with the control group, the Frontal_Mid_Orb_R, Occipital_Inf_R, and Temporal_Pole_Mid_R in the BWTC group were enhanced, and Frontal_Mid_L and Supp_Motor_Area_R in the BWTC group were weakened after the intervention. This study further explored the potential correlation between anxiety, depression, and local brain activity after the intervention. Frontal_Mid_Orb_R was significantly positively correlated with depression and Frontal_Mid_L had a significant negative correlation with anxiety.

Yang et al. (2016) found that depression and anxiety can be alleviated by modulating the functional connectivity between the interior and insular cortices of the dorsal cingulate gyrus, suggesting that the regulation of anxiety and depression is closely related to the functional connectivity of the brain. In other words, anxiety and depression can be alleviated by improving the functional connectivity of the local brain regions. Gold et al. (2015) identified the insula, thalamus, striatum, cingulate cortex, and amygdala as brain regions associated with anxiety, whereas Etkin et al. (2011) identified the insula, paracingulate cortex, and medial prefrontal cortex as brain regions associated with emotional engagement and processing, especially fear and anxiety. The insula, prefrontal cortex, anterior cingulate cortex, and amygdala play important roles in the development, processing, and regulation of depression and anxiety. Therefore, BWTC exercise improves anxiety and depression mainly by modulating the functional connectivity and neural activity of related brain regions. This study found that BWTC may relieve anxiety and depression by regulating neural activity in the Frontal_Mid_L and Frontal_Mid_Orb_R, respectively.

This study also analyzed the reasons why BWTC effectively alleviated anxiety and depression among college students. During practice, participants in the BWTC group were required to cooperate closely with their minds. Previous studies have shown that the mind completes information transmission through bioelectricity accompanied by neural and humoral regulation, generates neuronal excitement, and places the cerebral cortex in a special state of excitation, thereby regulating psychological and physiological activities (Ray, 2004; Hassan et al., 2019; Tang et al., 2019). This study attributes this effect to the emphasis of BWTC on the oneness of the body and mind, which stimulated the brain regions of the participants to a certain extent. Such stimulation

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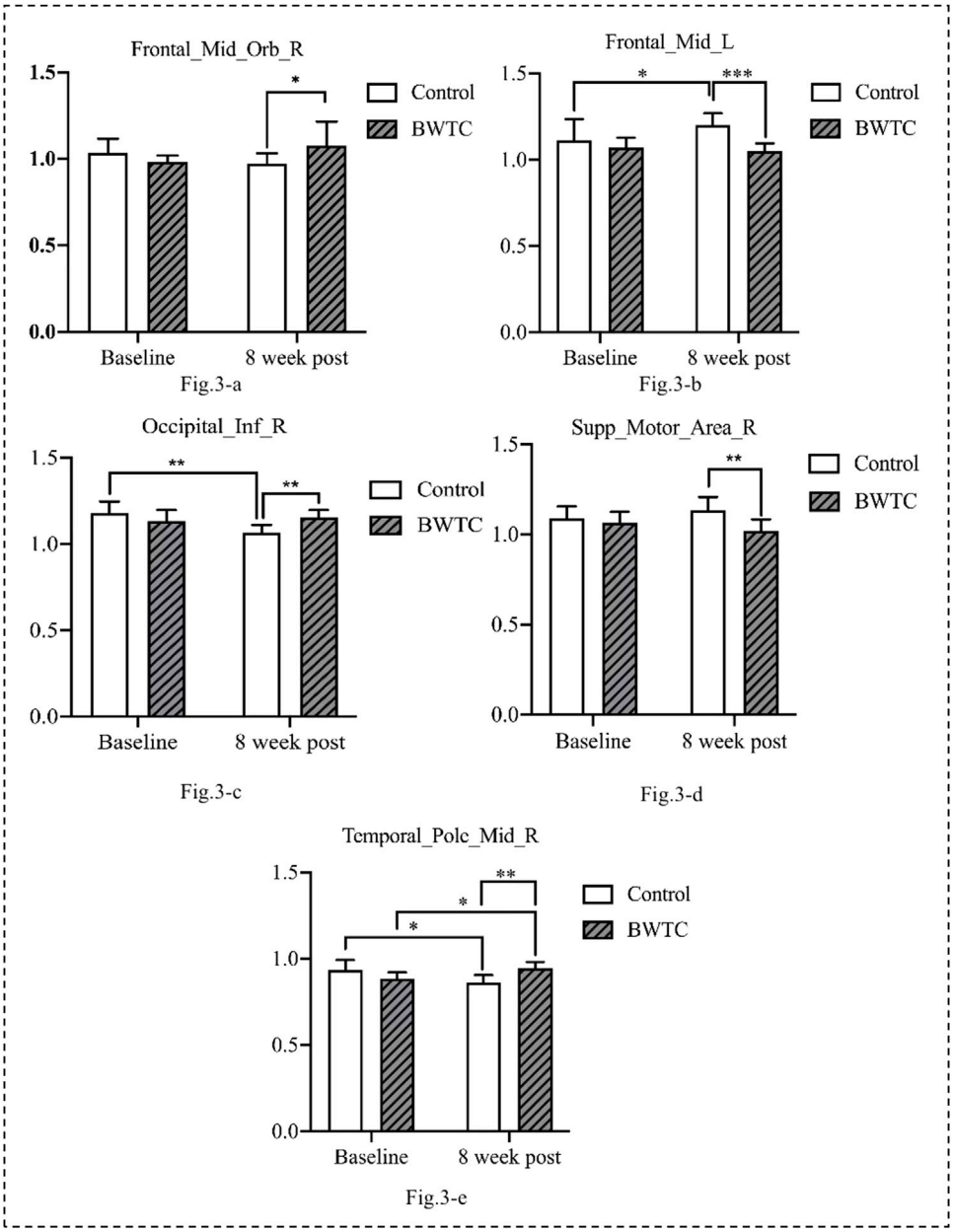

FIGURE 3
The differences between the groups before and after the intervention in the same brain region.

generates bioelectrical signals that lead to excitation and biochemical reactions, such as neurotransmitters and hormones, and activates key brain regions for emotion regulation, such as Frontal_Mid_L and Frontal_Mid_Orb_R. The brain network connections and mechanisms of these brain regions are optimized, thereby alleviating anxiety and depression in college students. BWTC is often accompanied by relaxing and soothing music, which can stimulate the auditory nervous system, relax tense nerves, and subsequently alleviate anxiety and depression (Petrovsky et al., 2015). A meta-analysis of 44 original studies by Song found that aerobic exercise may have a better effect on anxiety and stress, whereas traditional Chinese exercise may have a better effect on stress. This study found that BWTC might improve anxiety and depression.

This pilot study had several limitations. First, additional physical activities of the participants throughout the study could not be easily monitored. Second, the sample size was relatively small and the intervention was delivered over a relatively short period. Future

TABLE 5 Correlation between SAS-SDS and local brain activity after intervention.

Brain areas	SAS		SDS	
	r	*p*	r	*p*
Frontal_Mid_Orb_R	0.357	0.432	0.852	0.015
Frontal_Mid_L	−0.797	0.032	−0.445	0.317
Frontal_Sup_R	−0.214	0.645	−0.037	0.937
Insula_R	−0.250	0.589	0.482	0.274
Occipital_Inf_R	0.107	0.819	−0.148	0.751
Occipital_Sup_R	−0.179	0.702	−0.074	0.875
Supp_Motor_Area_R	−0.429	0.337	0.371	0.413
Temporal_Inf_L	0.214	0.645	0.036	0.757
Temporal_Pole_Mid_R	−0.357	0.432	−0.057	0.837

Note: SAS: self-rating anxiety scale; SDS: self-rating depression scale. Frontal_Mid_Orb_R: right middle frontal gyrus, orbital part; Frontal_Mid_L: left middle frontal gyrus; Frontal_Sup_R: right superior frontal gyrus; Insula_R: right insula; Occipital_Inf_R: right inferior occipital gyrus; Occipital_Sup_R: right superior occipital gyrus; Supp_Motor_Area_R: right supplementary motor area; Temporal_Inf_L: left inferior temporal gyrus; Temporal_Pole_Mid_R: right middle temporal gyrus of the temporal pole.

studies should employ larger samples and longer in-depth exercise durations to confirm the findings of this study.

5 Conclusion

In this pilot study with college students, BWTC had the potential to alleviate anxiety and depression in college students by regulating neural activity in Frontal_Mid_L and Frontal_Mid_Orb_R, respectively.

Data availability statement

The raw data supporting the conclusion of this article will be made available by the authors, without undue reservation.

Ethics statement

The studies involving human participants were reviewed and approved by the Experimental Ethics Committee of the Department of Psychology of Beijing Normal University and was conducted in compliance with the Declaration of Helsinki. The patients/participants provided their written informed consent to participate in this study.

Author contributions

SL and JZ conceived and designed the study protocol. The individual interviews were conducted by TG, ZS, YL, JZ, QW, and MC. JZ and TG analyzed the data. JZ, FP, QW, MC, ZC, and YL guided and supervised the Tai Chi training. JZ and SL contributed to the writing and reading of the manuscript. All authors gave their final approval and agreed to be accountable for all aspects of the work to ensure its integrity and accuracy.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Comparison and analysis of the biomechanics of the lower limbs of female tennis players of different levels in foot-up serve

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Purpose: The purpose of this study was to examine biomechanical performance of the foot-up serve (FUS) in female tennis players at different skill levels.

Methods: FUS analysis was completed in the biomechanical laboratory by 32 female college tennis players at three different levels. During FUS, 3D-biomechanical data from tennis players' lower limbs were collected. One-way ANOVA was used to examine differences in kinematic and kinetic data between groups

Results: Range of motion (ROM) of bilateral lower-limb joints revealed significant differences in kinematics performance during both the preparation and landing cushion phases ($p < 0.05$). During preparation, Level 3 was significantly longer than Level 2 (P-a = 0.042, P-b = 0.001, and P-c = 0.006). During the flight phase, significant differences between levels 1 and 3 (P-a:0.002) and levels 1 and 2 (P-c: 0.000) were discovered (P-a:0.002 and P-c:0.000). There were significant height differences between levels 1 and 2 as well as between levels 1 and 3. (P-a = 0.001, P-c = 0.000). During serve preparation (P-c = 0.001) and landing, GRF's peak was significantly higher than level 3. (P-c:0.007). Significant differences were found between groups in the LLS preparation stage, with level 3 significantly higher than levels 1 and 2. (P-a = 0.000, P-b = 0.001, and P-c = 0.000); during landing, level 2 LLS was significantly higher than levels 1 and 3. (P-a = 0.000, P-b = 0.000, and P-c = 0.035).

Conclusion: The range of motion of joints and the stiffness of the lower limbs have a significant impact on a tennis player's FUS performance. A larger of joint mobility and lower-limb stiffness promote better performance during the FUS preparation stage.

KEYWORDS

foot-up serve, women's tennis, biomechanics, motion control, lower limbs

1 Introduction

To compete at the highest levels in tennis, athletes must possess excellent speed, agility, muscle strength, and other abilities (Jaime et al., 2018). Athletes smash and serve regularly during tennis events in which serve is vital (Roetert et al., 2009). Thus, a powerful serve can easily win the match (Kovacs and Ellenbecker, 2011). Previous research has indicated that

the tennis serve is closely related to the upper limbs, but there have been little investigations on the lower limbs (Buckley and Kerwin, 1988; Elliott et al., 1995; Elliott et al., 2003). The lower limbs provide the foundation of the tennis service chain. Lower-limb movement is essential for energy generation and transmission to the trunk, upper limbs, and racket (Kibler, 1995).

Tennis players' lower limbs have a positive effect on how well their serves perform. Kovacs and Ellenbecker, (2011) claim that the tennis serve is a challenging motion that calls for athletes to coordinate their upper and lower limbs and joints in order to gather all forces generated on the ground and transfer them to the ball. Studies have shown a strong correlation between tennis serve speed and leg strength (Kaya et al., 2018). Reid et al. (2012) found that powerful performance of lower limbs can improve the racket's capacity to support speed. Olivier et al. (2005) also discovered that whereas tennis players at different levels of competition all demonstrate the same explosive force when serving, their vertical ground reaction force and coordination of lower limb connections greatly varies. These variations in biomechanical traits could affect how well athletes perform in competition. Therefore, it is crucial to conduct research on the lower limbs of the tennis serve to enhance performance.

Tennis serves are classified into two types based on the position of the feet when serving: foot-up serve (FUS) and foot-back serve (FBS). FBS supports both feet stably, which provides high balance and accurate service, but the force is relatively weak (Sun et al., 2012; Filippo et al., 2016). FUS is completed by relying on the support of the front leg and stepping up with the back leg. FUS has a higher hitting point and greater strength than FBS because of the forward movement of the center of gravity, although its balance is relatively low (Filippo et al., 2016; Liang et al., 2017). The majority of tennis players prefer to use FUS during a match. As a result, using biomechanical analysis technology to analyze the lower limbs during FUS serving not only improves understanding of the mechanism of tennis athletes' lower limbs but also has practical implications for tennis service training.

Here, the purpose of this study is to investigate lower-limb biomechanical performance in tennis players at multiple levels of performance during FUS using biomechanical analysis. China has combined athletic and academic endeavors to produce backup players for competitive sports since the 1980's (Guo et al., 2022), and the findings of this study will be useful for the training of both athletic and academic female tennis players. Based on previous research, we hypothesize that lower limb biomechanical performance differs significantly due to diverse level of female tennis players.

2 Research methods

A cross-sectional study with three parallel groups was used in this paper. Through comparing biomechanical indicators between three diverse level of female tennis players, we would like to figure out which biomechanical indicators would differ due to players' level.

2.1 Participants

A total of 32 female college tennis players participated in the test, and seven of them withdrew from the test due to test time, competition, and professional course conflicts. Finally, 25 female college tennis players, including seven national level 1 athletes (height: 178.0 ± 3.15 cm, weight: 58.5 ± 6.36 kg, and BMI: 18.46 ± 2.20 kg m^{-2}), eight national level 2 athletes (height: 174 ± 2.10 cm, weight: 56 ± 2.83 kg, and BMI: 18.06 ± 1.20 kg m^{-2}), and 10 students majored in the tennis special class (height: 168.5 ± 1.10 cm, weight: 54 ± 1.70 kg, and BMI: 19.01 ± 0.11 kg m^{-2}), participated in the biomechanical test for FUS. Criteria for subject recruitment were as follows: 1) the right hand of subject was the dominant hand; 2) subject had received professional tennis training for at least 1 year; 3) lower limbs and feet had no injuries or diseases in the past 6 months; and 4) participants had no other factors that could affect completing FUS.

2.2 Biomechanical tests

2.2.1 Biomechanical records

FUS test was performed out on the test participants in a laboratory. The kinematic and kinetic performance of the bilateral lower limbs during FUS was recorded using a 3D force platform (AMTI, United States of America) and an infrared Vicon motion capture system (Oxford Metrics Ltd, Oxford, UK). For kinematics, the frequency was 250 Hz, and for ground reaction force, it was 1000 Hz. Lower-limb stiffness (LLS), take-off height H), landing time (LT), ground reaction force (GRF), and lower-limb displacement (LLD) were the biomechanical indicators that were gathered. All data had been exported automatically by Vicon software.

2.2.2 Joint anatomical position

The kinematic data of both lower limbs of FUS were gathered in this study using the plug-in gait lower limb model (Figure 1). The experimental operators calibrated the athletes once they became accustomed to the experimental setting and constructed the three - dimensional image before the experiment. Then, using a ruler, the scientist completed the work required for personalized static modeling and measured the athletes' bilateral lower limbs' leg length, knee width, and ankle width. The lower limb joints of the athletes were then calibrated using reflecting markers by a scientist. The plug-in gait had 16 anatomical positions, including the left anterior superior spine (LASI), the right anterior superior spine (RASI), the left posterior superior spine (LPSI), the right posterior superior spine (RPSI), the left and right knees, the left and right tibias, the left and right ankles, the left and right toes, and the left and right heels (RTOE).

2.2.3 Action division of FUS

Tennis' service action could be divided into three stages: take-off before service, leaving, and landing cushioning, according to the literature (Harriss and Atkinson, 2009). In this study, the kinematics and dynamics of athletes' takeoff and landing cushions were primarily examined.

FIGURE 1
Lower-limb anatomy location in participants.

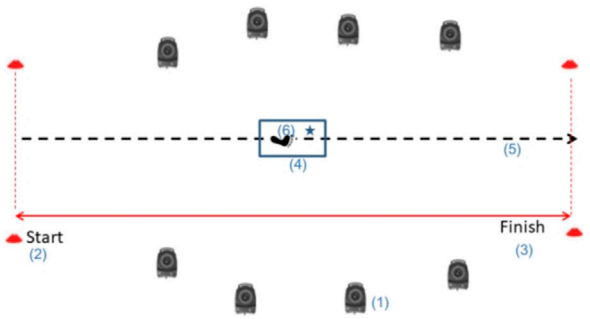

FIGURE 2
Experimental environment of FUS's biomechanical test.

2.3 Experimental protocol

Three steps made up the experimental test process: calibration prior to the experiment and familiarization of the athletes with the testing environment; formal experimentation; and data collecting and analysis following the experiment. The athletes had enough time to get used to the experimental test environment before the test. Following that, participants successfully completed six FUS with flat shot (Figure 2) on the force platform while maintaining consistency and coordination in their movements. During FUS, the athletes' bilateral lower limbs' hip, knee, and ankle joints' kinematic and kinetic data were gathered using a motion capture system. All participants were given 30 s to rest after each FUS to prevent fatigue. Throughout each test, the air conditioner maintained the test environment's room temperature at roughly 26 C. Following the test, the examiner standardized the subject's test results, and the participants engaged in relaxation exercises.

2.4 Data processing and analysis

2.4.1 Take-off height

The athletes' take-off height was calculated by using the take-off time, and the calculation formula was as follows: $height = \frac{(g \times \Delta t^2)}{8}$, g = 9.8 m/s, where Δt is the take-off time of athletes.

2.4.2 Calculation of LLS

FUS is mainly completed in the sagittal plane and by flexion of the knee joint. The hip joint and trunk remain relatively stable. Therefore, the LLS of athletes in the take-off preparation stage and landing r stage of FUS could be expressed by vertical stiffness. The calculation formula was as follows: $Kvert = \frac{Fmax}{\Delta y}$; where Fmax is the peak value of GRF, and Δy is the vertical displacement of the body's center of gravity. The vertical displacement could be calculated by the proportion of human morphological links (David, 2009), height, and angle of knee flexion extension ROM in the service preparation stage and the landing buffer stage. The calculation formula was as follows:

$$\Delta y = L_{thigh} - L \times \sqrt{L_{shank}^2 + L_{thigh}^2 - 2 \times L_{shank} \times L_{thigh} \times \cos\alpha},$$

$$L_{knee} = 0.039 \times H,\ L_{shank} = 0.285 \times H,\ L_{thigh} = 0.53 \times H.$$

2.4.3 Statistical analysis

The data were present as mean ± standard deviation (M±SD). Following the initial completion of standardization, the normal distribution of ROM, H, TOT, FT, LT, LLS, GRF, and LLD was tested (IBM, United States of America). One-way ANOVA was used to examine the differences between groups once the normal distribution test was finished. LSD *post hoc* analysis was used to analyze intergroup differences. P was set to 0.05. All analysis was performed using SPSS (IBM SPSS17.0, SPSS Inc).

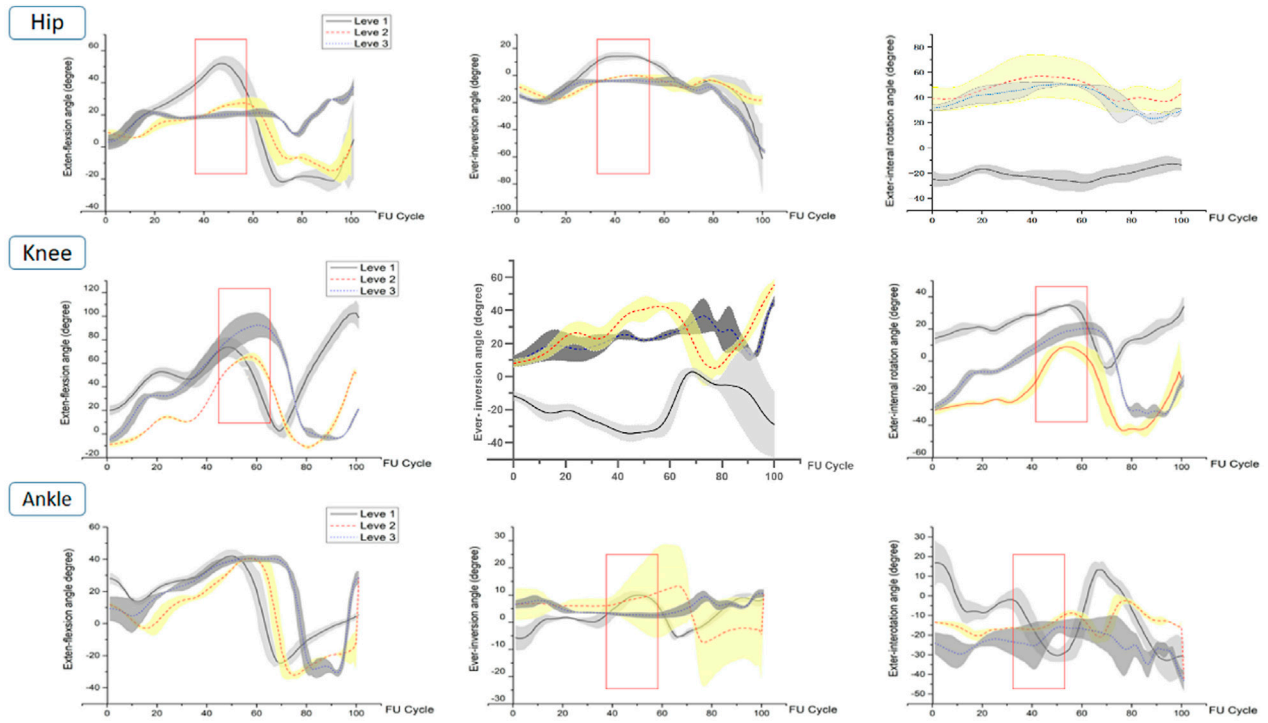

FIGURE 3
Right lower-limb joint ROM of level 1, level 2, and level 3 during FUS. The red rectangles represent the areas of difference in joint ROM during the take-off phase.

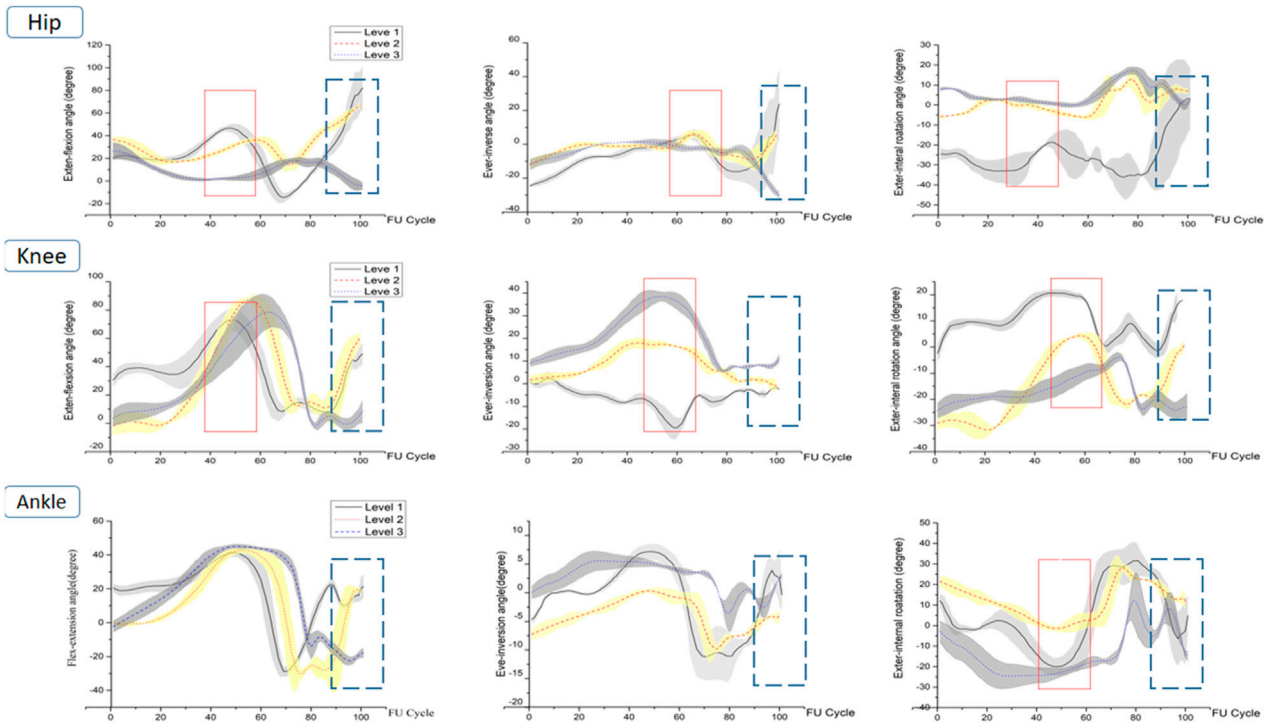

FIGURE 4
Left lower-limb joint ROM of level 1, level 2, and level 3 during FUS. (1) The red rectangles represent the areas of difference in joint ROM during the take-off preparation phase. (2) The blue dashed rectangles represent the areas of difference in joint ROM during the landing buffer phase.

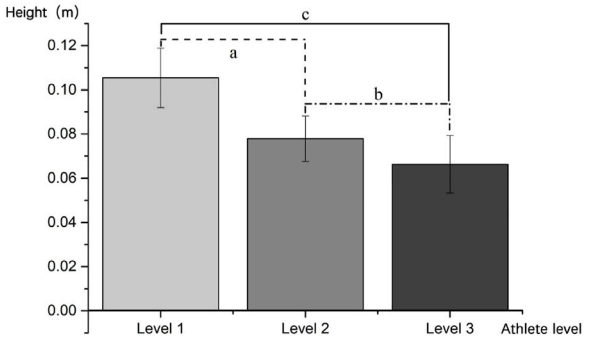

FIGURE 5
FUS's flying height of level 1, level 2, and level 3.

3 Results

3.1 Kinematic results

3.1.1 ROM of joints in FUS

Joint ROM performance of the bilateral lower-limb hip, knee, and ankle joints among groups during FUS preparation stage are shown in Figures 3, 4 and Table 1. Differences in the right lower-limb joints among three groups were found in hip and ankle joints. The flexion-extension, adduction-abduction, and internal rotation-external rotation ROMs of the hip were significantly greater in the level 1 and level 2 than in level 3 (P-a<0.001, P-b<0.001, and P-c = 0.001). The flexion-extension ROM in right lower-limb knee of level 1 was greater than the level 2 and level 3, however only ROM in knee of level 1 and level 2 was significantly different (P-a:0.002). Knee internal-external rotation ROM of Level 1 was significantly higher than level 3 (P-c: 0.016). The differences of ROM in ankle in the coronal and horizontal planes showed similar trends (level 1 > level 3 > level 2) (P-a:<0.001, P-c:<0.001).

During the preparation for FUS, significant differences in the hip and knee joints of the left lower limb were found (Figure 4; Table 2). Hip flexion-extension ROM was substantially larger at level 1 than at levels 2 and 3. Hip adduction-abduction ROM significantly increased with athletes' level (P-a:<0.001, P-b:0.014, P-c: <0.001). Knee flexion-extension ROM showed a level 2 > level 3 > level 1 trend across groups, however there was a significant difference between them. In terms of the ROM for knee internal-external rotation, level 2 was followed by levels 1 and 3. However there were significant differences between levels 1 and 2, Levels 2 and 3. Ankle internal-external ROM across levels 1 and 2, as well as between levels 2 and 3, differed significantly.

The hip joint's ROM during landing matched that of the FUS preparation stage, which was level 1 > level 2 > level 3. There were no differences between levels 2 and 3 in terms of internal-external rotation ROM or flexion-extension ROM, though. Similar trends were observed in the knee's flexion-extension ROM and internal-external rotation ROM; level 2 was significantly greater than level 1 and level 3, respectively (P-b:0.019, P-c:0.002; P-b:0.000, P-c: 0.000). Level 2 had a significantly larger adduction-abduction ROM of the knee than the other levels. Ankle flexion-extension ROM was considerably higher at levels 1 and 2 compared to level 3 with level 2 being higher than level 1. However, only the inversion-eversion ROM between level 1 and level 2, between level 2 and level 3, and the internal-external rotation ROM between levels 1 and 2 showed significant differences. The trend of ankle inversion-eversion ROM and internal-external rotation ROM among the three groups was level 1 > level 3 > level 2.

3.1.2 Time and flying height in FUS's different phases

The amount of time level 1, level 2, and level 3 participants took to complete FUS at various points varied significantly (Table 3). Level 3 was significantly longer than Level 2 during preparation (P-a = 0.042, P-b = 0.001, and P-c = 0.006). However, during flight phase, only the differences between level 1 and level 3 (P-a:0.002) and level 1 and level 2 (P-c:0.000) were significantly found (P-a:0.002 and P-c:0.000). LT of level 2 was much higher during landing than that of level 3 (P-b:0.035). Details of the FUS-H are shown in Table 4 and Figure 5 for levels 1, 2, and 3. Significant differences between level 1 and level 2 and between level 1 and level 3 were found in groups where the trend of H was level 1 > level 2 > level 3. (P-a:0.001, P-b:0.125, and P-c:0.000).

3.2 FUS-GRF

GRFs among three groups showed a bimodal peak force during serve preparation (Figure 6; Table 5). GRF`s peak of level 1 and level 2 were slightly higher than level 3; only significant difference between level 1 and level 3 was found (P-c:0.001). During landing, GRF peak of level 1 were significantly higher than level 3 (P-c:0.007), but there was no significant difference between level 1 and level 2 and between level 2 and level 3. The time of the GRF peak significantly moved forward with the increase in athlete level. (P-a:0.000 and P-c:0.000).

3.3 FUS-LLS

The differences in LLS during the landing and FUS preparation are shown in Table 6. Significant differences were seen in the preparation stage of LLS between groups, with level 3 significantly higher than level 1 and level 2 (leve3>level l > level 2). (P-a:0.000, P-b:0.001, and P-c:0.000). However, during landing, LLS of level 2 was much higher than that of levels 1 and 3. (P-a:0.000, P-b:0.000, and P-c:0.035).

4 Discussion

FUS has the advantage of hitting height and speed when compared to FBS. As a result, a thorough biomechanical analysis of FUS will help athletes improve their athletic performance. We compared the biomechanical performance of both lower limbs during FUS at levels 1, 2, and 3. Results showed that those who performed at a higher level had greater ROM throughout the FUS preparation stage but significantly different ROM in the coronal and horizontal planes of the knee and ankle, according to the results. The significant large joint ROM was used to land, but different levels of lower-limb joints take on different cushioning responsibilities

TABLE 1 Bilateral lower-limb joint ROM of level 1, level 2, and level 3 during the takeoff phase.

Joint	Level	Right lower-limb			Left lower-limb		
		ROM (deg)		*p*-value	ROM (deg)		*p*-value
		M	SD	P	M	SD	P
Hip joint	level 1	77.27[ac]	2.46	a:0.000	55.30[ac]	6.60	a:0.000
Flexion - extension	level 2	35.89[ab]	3.03	b:0.000	25.71[a]	2.20	c:0.000
	level 3	22.95[bc]	4.01	c:0.001	29.81[c]	4.97	a:0.000
	level 1	78.51ac	14.06	a:0.000	29.72[ac]	3.41	b:0.014
Adduction - abduction	level 2	17.97[a]	1.59	c:0.000	19.20 [ab]	2.53	c:0.000
	level 3	17.12[c]	3.24	a:0.001	13.48 [bc]	1.89	a:0.000
	level 1	39.85[ac]	3.74	c:0.002	17.61	0.70	b:0.019
Internal rotation - external rotation	level 2	28.62[a]	2.21	a:0.002	17.17	2.64	c:0.000
	level 3	28.33c	3.55	b:0.001	17.95	2.67	a:0.048
Knee joint	level 1	106.74[a]	7.31	c:0.001	61.14[ac]	3.15	b:0.000
Flexion - extension	level 2	88.14	5.95	c:0.016	93.31[ab]	2.16	c:0.001
	level 3	97.36	10.79	a:0.000	77.66[bc]	12.86	a:0.00
	level 1	43.28[c]	2.47	b:0.043	22.97[ac]	4.82	b:0.00
Adduction - abduction	level 2	43.42[b]	4.97	c:0.000	17.45[ab]	2.55	a:0.03
	level 3	31.58[c]	3.55	a:0.000	34.30[bc]	2.26	b:0.02
	level 1	44.59	5.12	c:0.000	21.59[ab]	3.33	
Internal rotation - external rotation	level 2	49.09	3.75		37.19[b]	1.73	
	level 3	52.73	3.75		20.02	4.07	
ankle joint	level 1	68.32	4.11		54.84	11.76	
Metatarsus flexion - dorsiflexion	level 2	71.24	4.35		67.21	16.34	
	level 3	62.99	10.34		63.55	1.75	
	level 1	18.54[ac]	3.00		15.45	4.06	
Inversion - eversion	level 2	4.80[a]	1.12		10.55	1.64	
	level 3	8.20[c]	1.52		9.92	4.20	
	level 1	57.57[ac]	10.59		43.61[a]	7.85	
Internal rotation - external rotation	level 2	17.66[a]	4.89		30.78[b]	3.79	
	level 3	26.54[c]	5.44		48.17	9.12	

during landing. Higher level participants demonstrated faster serve preparation times, longer take-off times, and more body displacement than lower level participants. The peak force point of GRF advanced gradually as athletes' performance improved during the preparation and landing phases. While FUS preparation improved subject performance, LLS demonstrated significantly different subject performance during the landing cushion. Different biomechanical performance across various performance levels during FUS point to the existence of a different dynamic movement chain, which would result in a different muscle working model and increased risk of injury and disease. Further research into the mechanisms and consequences behind this biomechanical performance is deserves.

4.1 Kinematic performance of FUS

Tennis players need to practice in frequent and consistent special training in order to achieve high-level sports performance and skilled movement abilities (Ben Kibler and Safran, 2000). Even young tennis players at a lower level require specialized training for an average of 6.1 days per week and 2.3 h per day to be in good competitive

TABLE 2 Left lower-limb joint ROM of level 1, level 2, and level 3 during the landing phase.

Joint	Level	Left lower-limb		
		ROM (deg)		*p*-value
		M	SD	P
Hip joint	level 1	46.86[ac]	14.03	a:0.004
Flexion - extension	level 2	20.60[a]	1.84	c:0.000
	level 3	6.07c	2.23	a:0.000
	level 1	38.31[ac]	3.54	b: 0.034
Adduction - abduction	level 2	16.48[ab]	1.17	c:0.000
	level 3	10.92[bc]	2.96	a:0.027
	level 1	18.98[ac]	8.96	c:0.009
Internal rotation - external rotation	level 2	7.18[a]	4.10	b: 0.000
	level 3	4.54[c]	1.18	c:0.002
Knee joint	level 1	36.60[b]	4.86	a:0.000
Flexion - extension	level 2	48.67[c]	10.54	b: 0.000
	level 3	11.19[bc]	4.77	b: 0.000
	level 1	2.90[a]	1.49	c:0.000
Adduction - abduction	level 2	5.02[c]	1.65	a:0.000
	level 3	3.74	2.05	b: 0.000
	level 1	19.05[b]	5.98	a:0.000
Internal rotation - external rotation	level 2	21.13[c]	4.00	c:0.004
	level 3	3.09[bc]	0.70	a:0.003
Ankle joint	level 1	11.43[a]	2.37	
Metatarsus flexion - dorsiflexion	level 2	49.15[ab]	6.99	
	level 3	6.04[b]	0.16	
	level 1	11.28[ac]	0.88	
Inversion - eversion	level 2	2.35[a]	1.79	
	level 3	5.11[b]	2.43	
	level 1	33.54[a]	2.34	
Internal rotation - external rotation	level 2	8.88[a]	4.33	
	level 3	21.41	10.05	

[a]Represented a significant difference between level 1 and level 2, $p < 0.05$.
[c]Represented a significant difference between level 2 and level 3, $p < 0.05$.
[b]Represented a significant difference between level 3 and level 1, $p < 0.05$.

condition (Kibler et al., 1988). Tennis movement skills, which are crucial for the formation and development of tennis athletes' movement abilities and the prevention of sports injuries, can be improved with the help of biomechanical theories and methods,

TABLE 3 Time spent in FUS's different phases of level 1, level 2, and level 3.

Phase	Level	Time (s)		*p*-value
		M	SD	
TOT				
	Level 1	0.90[ac]	0.07	a:0.042
	Level 2	1.16[ab]	0.07	b:0.000
	Level 3	1.55[bc]	0.37	c:0.006
FT				
	Level 1	0.29[ac]	0.02	a:0.002
	Level 2	0.25[a]	0.02	
	Level 3	0.23[b]	0.02	c:0.000
LT				
	Level 1	0.14	0.01	
	Level 2	0.16[c]	0.06	b:0.035
	Level 3	0.12[c]	0.02	

[a]Represented a significant difference between level 1 and level 2, $p < 0.05$.
[b]Represented a significant difference between level 3 and level 1, $p < 0.05$.
[c]Represented a significant difference between level 2 and level 3, $p < 0.05$.

TABLE 4 FUS's flying height of level 1, level 2, and level 3.

Level	FT(s)	H(m)	*p*-Value
Level 1	0.29 ± 0.02	0.11 ± 0.01[ac]	a:0.001
Level 2	0.25 ± 0.02	0.08 ± 0.01[a]	b:0.125
Level 3	0.23 ± 0.02	0.07 ± 0.01[c]	c:0.000

[a]Represented a significant difference between level 1 and level 2, $p < 0.05$.
[b]Represented a significant difference between level 2 and level 3, $p < 0.05$.
[c]Represented a significant difference between level 3 and level 1, $p < 0.05$.

which can be used by coaches, tennis players, and scientists (Elliott, 2006). Therefore, to enhance tennis performance and avoid injuries, biomechanical research techniques are required to understand the process of each stroke.

In this study, the ROM of bilateral lower limb joints of participants with higher performance level were greater than that of participants with lower performance level, especially the ROM of the hip joint of the left leg and the ROM of the hip joint, knee joint, and ankle joint of the right leg. As seen by previous studies, tennis players with a large ROM can better activate muscles during sports, allowing them to achieve greater take-off strength (Hu, 2009). Tennis service requires coordination as well as explosive acceleration ability of trunk and lower limb muscles (Liu, 2008). Higher levels of lower-limb strength and efficiency during take-off serve can provide a foundation for athletes' serve accuracy and lower-limb and trunk explosive force from movement technology (Yan et al., 2000; Jin and Qu, 2008). Therefore, the athletes' lower limb extensor muscle groups' eccentric contraction time and work distance were extended due to the large ROM of the joint prior to hitting the ball, allowing them to conserve a greater amount of elastic potential energy at this point. This larger

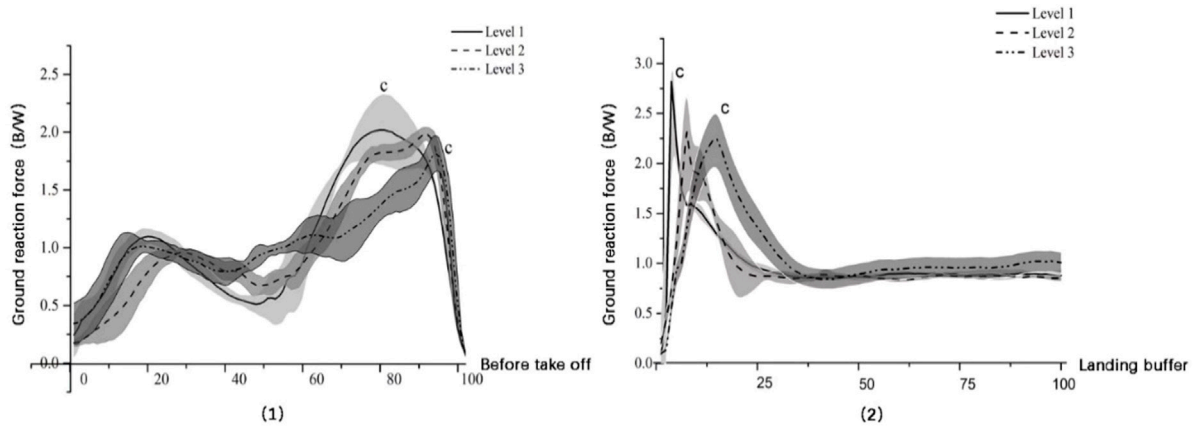

FIGURE 6
FUS's GRF variations in take-off (Right) and landing (Left) phases.

TABLE 5 Timing characteristic of GRF in level 1, level 2, and level 3 during takeoff and landing phases.

Level	GRF- peak value (B/W)	*p*-value	Stage of emergence (%)	*p*-value
Before take-off				
Level 1	2.02 ± 0.32[c]	a:0.117	76.61 ± 2.68[ac]	a:0.000
Level 2	1.99 ± 0.22	b:0.056	90.39 ± 0.82[a]	b:0.075
Level 3	1.81 ± 0.32[c]	c:0.001	92.65 ± 1.49[c]	c:0.000
Landing phases				
Level 1	2.81 ± 0.33[c]	a:0.127	2.29 ± 0.51[ab]	a:0.000
Level 2	2.31 ± 0.55	b:0.152	6.54 ± 1.19[bc]	b:0.000
Level 3	2.30 ± 0.19[c]	c:0.007	11.44 ± 1.34[ac]	c:0.000

[a]Represented a significant difference between level 1 and level 2, $p < 0.05$.
[b]Represented a significant difference between level 2 and level 3, $p < 0.05$.
[c]Represented a significant difference between level 3 and level 1, $p < 0.05$.

TABLE 6 Lower-limb stiffness of level 1, level 2, and level 3 during FUS.

Level	GRF- peak value (B/W)	LLD (cm)	LLS	P
Before take-off				
Level 1	2.02 ± 0.32	34.7 ± 1.50	5.81 ± 3.76[ab]	a:0.000
Level 2	1.99 ± 0.22	64.2 ± 0.29	3.09 ± 0.65[bc]	b:0.001
Level 3	1.81 ± 0.32	16.87 ± 0.67	10.92 ± 0.36[ac]	c:0.000
Landing phases				
Level 1	2.81 ± 0.33	45.22 ± 0.32	6.25 ± 0.72[ab]	a:0.000
Level 2	2.31 ± 0.55	25.52 ± 0.31	9.80 ± 1.19[bc]	b:0.000
Level 3	2.30 ± 0.19	30.64 ± 0.19	7.42 ± 0.62[ac]	c:0.035

[a]represented a significant difference between level 1 and level 2, $p < 0.05$.
[b]Represented a significant difference between level 2 and level 3, $p < 0.05$.
[c]Represented a significant difference between level 3 and level 1, $p < 0.05$.

ROM of the joint during the preparation stage of FUS is more conducive to the athletes' serve. However, an excessively large ROM may raise the burden of the lower limb extensors during training, which is unfavorable to the lower-limb takeoff during service. Additionally, it would have an impact on the coordination of the lower limb connections and the angular velocity of joints, which would have an impact on how well muscles store elastic potential energy and produce force.

During tennis serve, participants perform the landing by changing the ROM of joints. However, this ROM motion is not an independent activity, and it is accompanied with changes in ROM of adjacent joints (Chun-Man et al., 2011). Our results are supporting this finding. We also found that participants mainly relied on the left lower limb for landing cushion. Level 1 was conducted through ROM of hip and ankle joints, level 2 was conducted through ROM of the knee joint, and level 3 was relatively insignificant conducted through lower-limb ROM. These results are in line with research of Chun-Man et al. (2011) and Blackburn and Pauda (2008), which tennis player cushion the impact of landing by increasing the ROM of lower limb joints. These suggest that tennis service mainly depends on the athletes' single lower limb to perform the landing, the athletes' single lower limb needs to buffer or transfer the energy generated during the takeoff preparation of both lower limbs. This cushion strategy during landing is similar to the landing cushioning mechanism of volleyball athletes, which may lead player into a high risk of injury (Boling and Padua, 2013).

Studies on the control of lower limbs upon landing are critical for avoiding sports injuries. According to research, there is a clear relationship between the function of the hip and knee joints and the changes in the lower limb mechanism during landing (Boling and Padua, 2013). Power (2010) investigated the relationship between abnormal mechanical hip joint performance and knee joint injury and discovered that weak hip muscle force increases the degree of flexion of the ipsilateral trunk at the landing stage and increases the load of knee joint abduction. The increased muscle strength of the hip muscle group not only improves control of the athletes' lower limb movements during the landing phase and reduces the load on the knee and ankle joints during the landing phase, but it also supports the athletes' trunk stability during the landing phase. As a result, we proposed that athletes be encouraged to use the muscles surrounding the hip joint to cushion the landing.

4.2 Spatial performance and GRF of FUS

FUS may be separated into three stages: preparation, flight, and landing buffer based on the force distribution of the foot during serve. When preparing for takeoff, TOT of individuals with greater athletic levels was shorter than that of their competitors. Higher exercise levels were associated with longer FT and higher H during the flight phase, whereas lower performance levels were associated with the inverse. We also discovered that during the FUS preparation and landing cushion phases, the peak of GRF at level 1 was substantially larger than that at level 3. The force between foot and ground has been the source of FUS's strength. The GRF peak value, peak time phase, and action time all had a significant impact on impulse change. As a result, the law of conservation of impulse and momentum could be applied. Classical mechanics defines impulse as a process that involves the accumulation of force and time on objects as well as the increase in momentum (Yang et al., 2014). It is typically impacted by variables like quality, peak force, timing, and instantaneous load rate. More momentum can be gained in sports by people with greater sports levels than by those with lower sports levels. We discovered that the acquisition of impulse and the accumulation of momentum of athletes during the preparation stage of FUS had a more significant impact on the performance of the subsequent takeoff phase based on the characteristics of GRF and the temporal and geographical parameters of FUS. As a result, the FUS preparation phase warrants special attention from coaches and sports scientists.

4.3 Lower-limb stiffness performance in FUS

LLS of tennis players is related to their explosive output ability, according to Durand et al. (2010), and the higher LLS level, the higher reverse jump height. In the triple jump, Rabita et al. (2008) discovered that the LLS level of triple jumpers with enhanced training experience is significantly higher than that of jumpers without such training experience. Hobara et al. (2010) investigated LLS in endurance athletes and found similar results to Rabita et al. (2008). The biomechanical study of tennis also found that the LLS of athletes have a special impact on the efficiency of foot movement and help athletes achieve higher sports performance (Durand et al., 2010). Appropriate LLS can stimulate the degree of muscle activation to obtain a higher level of GRF and motion impulse through ROM (Harriss and Atkinson, 2009). This experiment discovered that the LLS of level 3 was greater than that of levels 1 and 2. The higher LLS at level 3 was primarily due to a smaller ROM, which LLS could not compensate for in terms of FUS performance. Bouhlel et al. (2006) discovered that moderate intensity exercise training can significantly improve the LLS of young male athletes' lower limbs. Hobara et al. (2008) discovered that different types of exercise can improve the LLS of participants' lower limbs. FUS is an explosive action, so explosive and enhanced exercise may be appropriate for tennis players when combined with sport-specific characteristics (Harrison et al., 2004). LLS is linked to sports injuries. In general, too high LLS can easily cause bone tissue structure damage, whereas too low LLS can easily cause soft tissue structure damage (Butler et al., 2003). Grimston et al. (1991) and Radin et al. (1978) discovered that excessive peak values of force, exercise loading, and impact loading put athletes at risk of injury and bone tissue damage.

Therefore, athletes with higher levels GRF peaks in the serve preparation stage would result in a higher risk of injury than athletes with lower levels. In addition, we also found that LLS of level 2 and level 3 was much higher than that of level 1 in the landing cushion phase. The kinematic analysis results showed that those with higher sports level had larger ROM than those with lower sports level in the landing cushion phase. Zhang et al. (2000) and Devita and Skelly (1992) found that increased knee ROM of joint flexion and extension to reduce moment of joint in the landing cushion phase can lead to reduced joint stiffness. Therefore, when coaches discover that athletes use smaller ROM to land, they should timely limit the movement mode or prompt athletes to land by increasing the ROM of sagittal joints, so as to avoid the occurrence of sports injury caused by exorbitant LLS.

Kinematic and kinetic analyses supported the hypothesis of this study, and revealed significantly different lower extremity biomechanical performance in tennis players serving at various sport levels; thus, real-time monitoring of lower extremity biomechanical performance during

serving would not only help to improve tennis serving performance, but would also provide athletes with a guarantee for reduced sports injury risk. However, the study has some limitations. For example, the study did not analyze surface electromyography (sEMG) in the experimental test; the analysis of joint working patterns during the serve may have some limitations; and it is necessary to include sEMG in the follow-up study to comprehensively understand the working patterns of lower limb muscles during the serve. Second, the study did not record the information of sports equipment in detail, such as shoes and racket types; this imperfect recording of may also have an impact on the understanding of the participants' basic characteristics and movement characteristics; thus, in order to improve the in-depth understanding of the study results, the information of the participants' equipment should be recorded in the follow-up study. The final limitation is the small sample size. Despite the fact that our study's sample size is in line with previous studies, this small size may not provide adequate statistical power. I Increasing the sample size might provide more insights into understanding of the biomechanical outcomes of tennis serves in future studies.

5 Conclusion

Women tennis players with varying levels of performance exhibit significantly different lower-limb biomechanical performance. The ROM of joints and the stiffness of the lower limbs have a major impact on a tennis player's FUS performance. Better performance during the FUS preparation stage is facilitated by a greater range of joint mobility and stiffer lower limbs. The lower-limb stiffness, however, has an impact on the tennis player's risk of injury during the landing cushion. Therefore, we propose that real-time player monitoring is necessary to assist different level women tennis players in enhancing their FUS performance.

Data availability statement

The original contributions presented in the study are included in the article/supplementary material, further inquiries can be directed to the corresponding author.

Ethics statement

The studies involving human participants were reviewed and approved by the ethics committee of Research Academy of Grand Health, Ningbo University. The patients/participants provided their written informed consent to participate in this study.

Author contributions

ZL and JW conceived the presented idea, developed the framework, and wrote the manuscript. SY, ZL, and YZ were involved in the data collection. JY, YG, and JL. Provided critical feedback and contributed to the final version. All authors have read and agreed to the published version of the manuscript.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

The reviewer YH declared a past co-authorship with the authors YG to the handling editor.

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The role and regulation mechanism of Chinese traditional fitness exercises on the bone and cartilage tissue in patients with osteoporosis: A narrative review

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Osteoporosis (ops) is a systemic degenerative bone disease characterized by bone mass reduction, bone mineral density loss, bone microstructure destruction, bone fragility, and increased fracture susceptibility. Thus far, drug therapy is the main method used to prevent and treat osteoporosis. However, long-term drug treatment will inevitably lead to drug resistance and certain side effects. In response, rehabilitation treatment is generally recommended, which involves drug supplementation combined with the treatment. A Chinese traditional fitness exercise is an organic combination of sports and traditional Chinese medicine with a series of advantages such as being safe, convenient, non-toxic, and harmless. Hence, it is one of the rehabilitation methods widely used in clinical practice. By searching the CNKI, PubMed, Web of Science, Embase, Cochrane Library, and other relevant databases, our research clarifies the current situation of four kinds of Chinese traditional fitness exercises widely used in clinical practice, namely, Taijiquan, Baduanjin, Wuqinxi, and Yijin Jing. In addition, the molecular mechanism of osteoporosis is summarized in this study. Based on the research, Chinese traditional fitness exercises are expected to directly stimulate the bone through a mechanical load to improve bone density. Moderate and regular traditional Chinese fitness exercises also improve osteoporosis by regulating the endocrine system with the secretion of hormones and factors such as estrogen and irisin, which are beneficial for bone formation. Finally, the purpose of promoting bone formation, reducing bone loss, and preventing and treating osteoporosis is achieved. The various means of Chinese traditional fitness exercises have different emphases, and the effect of improving bone density differs in various parts of the body. The exercisers may choose the exercise flexibly based on their own needs. Chinese traditional fitness exercises can improve the bone density of the exercisers and relieve pain, improve balance, and regulate the psychological state. Consequently, it is worth promoting to be applied in clinical practices.

KEYWORDS

Chinese traditional fitness exercise, osteoporosis, bone metabolism, bone mineral density, therapy

1 Introduction

Osteoporosis (ops) is a systemic degenerative bone disease characterized by bone mass reduction, bone density loss, bone microarchitecture destruction, bone fragility, and increased fracture susceptibility (Lane et al., 2000). Osteoporosis may occur in all genders and at any age, primarily in postmenopausal women and elderly people. In America, 10.2 million people suffer from osteoporosis and another 43.4 million are placed at an increased risk for fractures because of the low bone mass density (Arceo-Mendoza and Camacho, 2021). Meanwhile, in China, the incidence of osteoporosis is 51.6% and 10.7% in women and men above 60 years old, respectively (Ke et al., 2019). Approximately, 9 million new cases of osteoporosis are annually diagnosed worldwide. Thus, osteoporosis and its complications severely influence people's quality of life and cause a burden to the family and society. At present, drug and rehabilitation therapies are used as common methods to prevent and treat osteoporosis. Drug treatment falls into two categories. One is using drugs to adjust the metabolism of elderly osteoporosis patients. For example, people with senile osteoporosis take supplements for the deficiency of bone calcium and certain vitamins in the body by taking a certain amount of calcium and vitamin preparations. The other is using sex hormones to stimulate bone formation to reduce bone decomposition and achieve the goal of osteoporosis treatment (Reid and Billington, 2022). The sex hormone is effective in treating osteoporosis in postmenopausal women (Gosset et al., 2021). However, long-term drug treatment will inevitably lead to drug dependence and increase the risk of side effects. Drugs used to treat osteoporosis often need to be broken down and absorbed through the digestive tract because these drugs need to be taken for a long time. Therefore, it inevitably damages the gastrointestinal mucosa. Certain osteoporosis drugs may also cause transient head pain, insomnia, anxiety, and other symptoms; female patients with osteoporosis often need to take supplement estrogen drugs such as conjugated estrogen tablets. Moreover, the long-term use of osteoporosis drugs can cause patients to become prone to endometrial hyperplasia, thereby leading to malignant tumors such as endometrial cancer. Therefore, clinical rehabilitation treatment is generally recommended as a supplement combined with drug treatment as an alternative to medical treatment. Rehabilitation is a form of comprehensive treatment, including psychotherapy, dietary guidance, physical factor therapy, and exercise therapy. Psychotherapy helps patients to overcome and accept their illnesses while improving patients' compliance and initiative during the treatment. Scientific dietary guidance effectively helps treat patients with osteoporosis by supplementing Ca^{2+} and other trace elements related to bone formation directly and indirectly in the occurrence and development of osteoporosis. Physical factor therapy covers a wide range of therapies, including whole-body vibrations, ultrasounds, phototherapy, alternating current, direct current, magnetotherapy, cold and heat therapy, and fumigation and washing, which provides holistic treatment for osteoporosis-related aspects from different aspects. In addition, its effects include the stimulation and strengthening of muscles, bones, and ligaments; supplementing trace elements; counseling patients with negative emotions; and preventing and treating other potential complications. However, long-term comprehensive rehabilitation in the hospital may produce negative emotions for the patients and a great economic burden to their families. It will also take up more medical resources, thereby initiating the problems of the distribution of social medical resources.

Exercise therapy is very important. In 2018, the Chinese guidelines for the treatment of osteoporosis pointed out that exercise could improve body agility, strength, posture, and balance and effectively reduce the risk of falls and fractures caused by the diminishing bone mineral density. Therefore, exercise therapy is important to prevent and treat osteoporosis (Ma et al., 2019). Chinese traditional fitness exercises are aerobic exercises, guided by medical theories; it pays attention to the unity of mind, breath regulation, and movement. Moreover, it is the product of the organic integration of guiding techniques, such as qigong, martial arts, and medicine (Jahnke et al., 2010; Yeung et al., 2018; Yang et al., 2022). Regardless of the kind of traditional Chinese fitness exercise, it is based on the theories of yin and yang, zang-fu organs, qi and blood, meridians, and collaterals of traditional Chinese medicine. It takes nourishing essence, qi, and regulating spirit as the basic points of exercise, a dynamic form as the basic form of exercise, and guides the virtual, solid, dynamic, and static exercise with the theory of yin and yang. Flexion and pitching guided by opening and closing explain the harmony and unity of the form, spirit, qi, blood, surface and inner parts in sports, and fitness with the whole concept. Chinese traditional fitness exercises are clinically widely accepted by doctors, patients, and family members. They have various forms that are safe and easy to learn, create strong interest, and have almost no side effects. Taijiquan, Baduanjin, Wuqinxi, and Yijin Jing are widely used in treating osteoporosis. Taijiquan is light and soft; it combines movement and static, emphasizes the control of a breathing rhythm with consciousness, forms a linkage with action, improves the elasticity of joints and ligaments, strengthens muscle strength, promotes calcium deposition, prevents joint deformation and other diseases, and regulates the function of the viscera as a whole, which warms and takes care for the kidney and spleen. Meanwhile, Baduanjin's exercise movement is simple and slow, and it is primarily based on an isometric contraction of muscles. It emphasizes the training of the lower limb muscle strength to improve the balancing ability of the lower limbs to prevent patients from falling and reduce the occurrence of fractures. Furthermore, training for Baduanjin increases bone stress, promotes bone formation, reduces bone resorption, and controls the progress of osteoporosis. In addition, the movements of Wuqinxi are simple and safe, and the active parts are comprehensive. Its movements involve the major muscle groups of the whole body and the joint movements of the spine, limbs, and fingers, which may improve the blood circulation of the soft tissue of the spine and other joints, help maintain the normal structure of bones and joints, prevent joint stiffness, enhance muscle strength, and delay the development of osteoporosis. Meanwhile, Yijin Jing focuses on strengthening muscles and bones. Long-term practice can improve the flexibility of muscles and ligaments, increase bone density, and also, open blood vessels to prevent and treat diseases.

A comprehensive search was performed using PubMed, CNKI, Embase, Web of Science, MEDLINE, CBM, and Cochrane Library databases. The studies were published from the database inception to January 2023. Medical subject headings (MESH), keywords, and

free words were used in the retrieval strategy including "osteoporosis," "bone losses," "post-traumatic osteoporosis," "senile osteoporosis," "age-related osteoporosis," "involutional osteoporosis," "postmenopausal osteoporosis," "low bone mineral density," "Tai-ji," "Tai Chi," "Tai Ji Quan," "Taiji," "Taijiquan," "Tai Chi Chuan," "Baduanjin," "Wuqinxi," "Yijinjing," and "Chinese exercise." We collected all available RCTs of traditional Chinese fitness exercises on osteoporosis patients. The languages are limited to Chinese and English; the publication status is unrestricted. Exclusion criteria were as follows: 1) duplicate papers; 2) articles published in the abstract form or with incomplete data or when complete data could not be obtained after contacting the authors; and 3) no literature reports on the outcome measures. The current situation and the clinical application of four kinds of Chinese traditional fitness exercises including Taijiquan, Baduanjin, Wuqinxi, and Yijin Jing in osteoporosis are summarized in this review. The molecular mechanism is also briefly given.

2 The mechanism of Chinese traditional fitness exercises in improving osteoporosis

Exercise improves osteoporosis. Exercises can change the bone mass, morphology, and structure based on a mechanical load. It also affects bone metabolism by regulating the endocrine system to achieve the ultimate goal of preventing and treating osteoporosis.

The mechanical load generated by exercise acts on the bone tissue of the human body in two ways, directly stimulating the bone and indirectly affecting the bone through muscle contractions (Huang et al., 2008). The direct stimulation of mechanical loads indicates that the external force produced by exercise acts directly on the bone tissue. The bone tissue converts the changes of external forces during exercise into chemical or electrical signals. On one hand, exercise promotes the differentiation and proliferation of osteocytes to inhibit bone resorption, promote bone formation, and reduce bone mass loss. On the other hand, the bone trabeculae will be rearranged in bone remodeling during exercise, which could affect bone remodeling and change the structure and shape of the bone. Exercise improves bone strength and density, enhances the bone's mechanical load-bearing capacity, and prevents the occurrence of osteoporotic fractures (Marques et al., 2011). Indirect stimulation of muscle contractions during exercise indicates that it could stimulate the bone tissue and produce indirect stress such as tension, shear force, and a squeezing pressure on the bone. The traction not only produces benign stimulation to the bone quality but also improves the blood supply to the bone, thereby increasing the inorganic salt deposition in the blood such as calcium and phosphorus. Bone mineral density also increases, which helps in decelerating the loss of the bone mass (Kawano et al., 2002; He et al., 2003).

In addition to some clinical studies showing that exercise has certain effects on osteoporosis treatment, the cellular and molecular mechanisms of exercise against osteoporosis are better understood in animal and human cell models. The stimulation of OPG/RANKL, Wnt3a/β-catenin, NF-κB, and MAPK pathways plays an important role in the progression of osteoporosis. The level of OPG increases with whole-body vibrations or treadmill stimulations in osteoporotic rats. Osteoporosis could be effectively improved by regulating the OPG/RANKL pathway by decreasing the expression of RANKL and increasing the ratio of OPG/RANKL (Tang et al., 2006; Pichler et al., 2013; Notomi et al., 2014; Yuan et al., 2016). Downhill running can promote osteoblast differentiation and osteogenic abilities to enhance bone formation and metabolism. It improves the bone morphology and structure by activating the suppressed Wnt3a/β-catenin pathway in the bones of type 2 diabetic mice (Chen et al., 2021). In addition, the LRP family proteins and Wnt/β-catenin pathway are regulated by exercise, which improves bone remodeling, bone mass recovery, and bone stiffness in rats with disuse osteoporosis (Chen et al., 2016; Jia et al., 2019). Roller exercises prevent osteoporosis by activating CD8 T cells to release IFN-γ, which exhibits the effects of inhibiting osteoclast formation and recovering bone loss through NF-κB and MAPK pathways (Shen et al., 2022) (Figure 1).

Exercise can regulate bone metabolism by regulating hormone secretion in the endocrine system. Estrogen, muscle hormone, glucocorticoid, and parathyroid hormones participated in the process of bone metabolism (Gennari et al., 2010; Gardinier et al., 2015; Tong et al., 2019). Estrogen also promoted the proliferation and differentiation of osteoblasts while inhibiting the activity of osteoclasts and inducing osteoclast apoptosis (Robinson et al., 2009; Thouverey and Caverzasio, 2015; Lee et al., 2018; Gavali et al., 2019; Cline-Smith et al., 2020). In addition, estrogen plays a crucial role in maintaining the balance between bone resorption and bone formation to protect bone tissues. The abnormal regulation of the kynurenine pathway after estrogen deficiency may lead to the decrease of KAT expression and the concentration of KYNA of the serum in the muscles. The treadmill exercise increased the expression of the KAT enzyme in skeletal muscles while increasing the serum KYNA concentration, thereby reducing NF-κB phosphorylation by activating Gpr35. Thus, the treadmill exercise could significantly alleviate bone loss and bone microstructure destruction by inhibiting osteoclast maturation and osteoblast differentiation (Gavali et al., 2019). The iris induced by exercises can directly target osteoblasts and promote the proliferation, differentiation, and mineralization through the p38/ERK MAPK signal pathway (Qiao et al., 2016). The iris activated autophagy through the Wnt/β-catenin signal pathway and promoted osteogenic differentiation of bone marrow mesenchymal stem cells (Chen et al., 2020). Glucocorticoids negatively affect the progression of osteoporosis. Glucocorticoids inhibited the expression of Wnt16, which is an osteogenic signal to reduce the differentiation and maturation of osteoblasts by increasing the expression of Wnt pathway inhibitors such as sFRPs, WIF1, SOST, and Dickkopf-1 (DKK1) (Lane, 2019). Moreover, glucocorticoids can upregulate the expression of various factors that promote the differentiation of bone marrow mesenchymal stem cells into adipocytes, thereby decreasing the differentiation from bone marrow mesenchymal stem cells to osteoblasts (Han et al., 2019). Glucocorticoids also promoted osteoclast differentiation and maturation while extending the osteoclast life span and promoting bone resorption (Conaway et al., 2016). Exercise significantly increased blood glucocorticoid levels, which was closely related to the intensity of exercise but less related to the total amount and duration of exercise. Short-term high-intensity cycling exercises significantly increased the serum

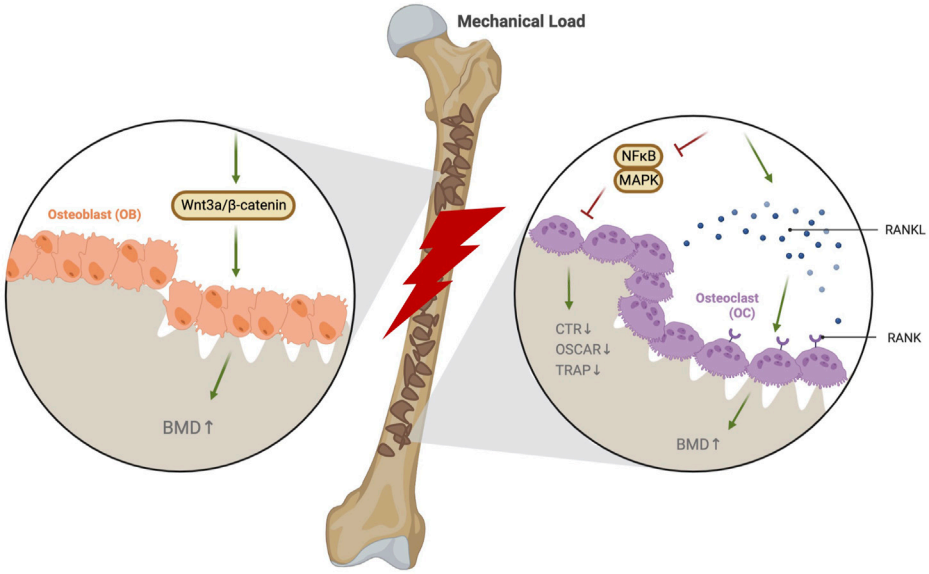

FIGURE 1
Exercise stimulates bone tissue through the mechanical load to enhance bone density.

cortisol concentration in the subjects (Córdova A et al., 2019). Thus, a scientific diet and regular exercise of the appropriate intensity were strongly recommended as effective means to treat glucocorticoid-induced osteoporosis in the latest "Prevention and Treatment of Glucocorticoid-Induced Osteoporosis in Adults: Consensus Recommendations from the Belgian Bone Club" published in 2022 (Laurent et al., 2022). On the other hand, the effects of the parathyroid hormone on bone metabolism were more complex. In maintaining bone homeostasis, the parathyroid hormone played a dual role in regulating the bone metabolism primarily based on Gs-cAMP/PKA, Gq/G11/PLC/PKC, non-PLC/PKC, and β-arrestin signal pathways (Lee and Lorenzo, 2002; Yang et al., 2007; Li et al., 2017; Li et al., 2018; Zhang et al., 2018). The continuous infusion of PTH increased bone resorption and bone loss; however, intermittent PTH therapy might stimulate bone formation by reactivating the bone surface and reducing osteoblast apoptosis (Nakajima et al., 2002; Rickard et al., 2006; Friedl et al., 2007). A significant increase was observed in the blood PTH within 4 h after a 60-min load-carrying treadmill exercise, which gradually returned to pre-exercise levels within 24 h after exercise (Staab et al., 2023). Therefore, regular exercise may increase PTH intermittently and eventually stimulate bone formation (Figure 2).

3 Chinese traditional fitness exercises for the prevention and treatment of osteoporosis

Researchers in different areas have confirmed various salutary functions of exercise in recent decades including improving osteoporosis. Exercise is important in increasing muscle flexibility and balance, increasing bone strength, slowing bone loss, and improving the bone structure. The guidelines for the rehabilitation of osteoporosis suggest that proper aerobic exercise functions by stimulating bone formation, regulating the endocrine system, increasing estrogen levels, promoting bone production, and preventing bone loss (Yuan and Wang, 2019). A Chinese traditional fitness exercise is a national traditional exercise with the concept of health preservation of traditional Chinese medicine combined with aerobic exercises. It is guided by "treating disease without disease" and is based on the principle of a "combination of movement and stillness, strength, and softness." Chinese traditional fitness exercises organically combine human physical activities, breathing, and mental regulation. Long-term exercise helps reduce bone loss and relieve pain, and effectively prevents and treats osteoporosis (Figure 3).

3.1 Chinese traditional fitness exercises increase bone mineral density and prevent bone loss

Chinese traditional fitness exercises effectively increased the bone mineral density and reduced bone mass loss in patients with osteoporosis. In the studies on the relationship between Taijiquan and osteoporosis, Taijiquan was intervened in elderly women with osteoporosis for 9 and 12 months. The results showed that Taijiquan effectively improved the bone mineral density of the lumbar spine, hip, femur, and forearm in elderly women with osteoporosis (Wayne et al., 2012; Wu et al., 2016). In addition, the bone mineral density of the spine and femur was effectively increased after practicing Baduanjin in patients with osteoporosis (Liu et al., 2015). Hence, it further concluded that the effect of combined drug therapy with Chinese traditional fitness exercises is significant. Similarly, 6 months of the Wuqinxi and Yijin Jing exercise achieved the same effect (Jing and Zhang, 2008; Shen et al., 2013; Shen et al., 2018).

TABLE 1 Different Chinese traditional fitness exercise types on the human OP.

Participant	Duration	Study group (*n*)	Outcome measures			Reference
			BMD	Bone metabolism	Pain	
Elderly women	48 weeks	Taijiquan (28), brisk walking (29), and square (32)	EG: lumbar spine↑ femoral neck↑			Song et al. (2018)
Menopausal women	10 months	Mulan boxing (12), Taijiquan (12), skipping (12), Taiji pushing hand (12), and control (12)	EG: lumbar spine↑ ulna↑ radius↑			Zhou (2004)
Perimenopausal women		Taijiquan (20) and control (20)	EG: lumbar spine↑ femoral neck↑	EG: BGP↓ ALP↓	EG: VAS↓	Song (2008)
Perimenopausal women	6 months	Taijiquan softball (15) and control (15)	EG: spine↑ the whole body; ↔ upper limb↔ pelvis↔	EG: Ca↔ P↔ Mg↔ ALP↔		Du et al. (2014)
Perimenopausal women	48 weeks	Taijiquan (36) and control (38)	EG: lumbar spine↔ femoral neck↔ large rotor↔ Ward's triangle↔			Zhao (2020)
Menopausal women		Taijiquan (31) and control (28)	EG: lumbar spine↑ femoral neck↑ Ward's triangle↑		EG: balance↑	Zou et al. (2011)
Community living elderly people	12 months	Taijiquan (58), resistance exercise (59), and control (59)	EG: lumbar spine↔ total hip↑			Woo et al. (2007)
Postmenopausal women	12 months	Taijiquan (65) and control (67)	EG: lumbar spine↔ proximal femur↔ distal tibia↑			Chan et al. (2004)
Postmenopausal women		Taijiquan (48) and control (51)	EG: lumbar spine↑ proximal femur↑ Ward's triangle↑			Qin et al. (2005)
Elderly females	12 months	Taijiquan (35), dance (35), and walking (35)	EG: calcaneus↑		EG: balance↑	Song et al. (2014)
Postmenopausal women	12 months	Taijiquan (17) and control (17)	EG: lumbar spine↑ femoral neck↑ distal tibia↑ ultradistal tibia↑			Qin et al. (2002)
Postmenopausal women	12 months	Traditional Taijiquan (40), simplified Taijiquan (40), and control (39)	EG: lumbar spine↑ femoral neck↑ Ward's triangle↑			Wang et al. (2015)
Adults	12 weeks	Taijiquan (129), walking (121), and control (124)	EG: lumbar spine↑ total hip↑			Hui et al. (2015)
Postmenopausal osteogenic women	9 months	Usual care(43) and Taijiquan (43)	EG: lumbar spine↑ Total hip↑ femoral neck↑	EG: CTX↓ OSC↓	EG: SF-36↔ MENQOL↔ PAR↔ balance↑	Wayne et al. (2012)
Elderly people	24 weeks	Taijiquan (14) and resistance training (14)		EG: PTH↑ urinary calcium↓ PYD ↔ BAP↑		Shen et al. (2007)
Postmenopausal women	6 months	Placebo + Taijiquan (37) and placebo (37)		EG: Ca↔ PTH↑ urinary calcium↔ P↔ BAP↑ BAP/TRAP↑		Shen et al. (2012)
Postmenopausal osteogenic women	6 months	Placebo + Taijiquan (37) and placebo (37)		EG: 8-OHdG↓ GTP↑		Qian et al. (2012)
Osteoporosis in elderly females	24 weeks	Taijiquan (30) and control (31)	EG: anklebone↑		EG: SF-36↑ balance↑	Chyu et al. (2010)
Osteoporosis in elderly females	6 months	Taijiquan (22) and control (22)			EG: SF-36↑ balance↑	Alp et al. (2009)
Community-dwelling females	12 weeks	Taijiquan (31)			EG: balance↑	Murphy and Singh (2008)

(Continued on following page)

TABLE 1 (*Continued*) Different Chinese traditional fitness exercise types on the human OP.

Participant	Duration	Study group (*n*)	Outcome measures			Reference
			BMD	Bone metabolism	Pain	
Osteoporosis in elderly males	18 weeks	Taijiquan (25) and control (24)			EG: balance↑	Maciaszek et al. (2007)
Healthy middle-aged females	12 weeks	Baduanjin (44) control (43)	EG: BMD↑	EG: IL-6↓		Chen et al. (2006)
Postmenopausal osteogenic females	12 months	Control (42), Ca (45), Baduanjin (48), and Baduanjin + Ca (49)	EG: lumbar spine↑ femoral neck↑		EG: VAS↓ balance↑	Liu et al. (2015)
Osteoporosis in elderly people	6 months	Baduanjin + Caltrate D (44) and Caltrate D (44)			EG: balance↑	Li et al. (2019)
Elderly females	24 weeks	Wuqinxi (36) and control (35)	EG: lumbar spine↔ femoral neck↔		EG: balance↑	Wang et al. (2018)
Patients with primary type I osteoporosis	6 months	Wuqinxi (30) and control (30)	EG: lumbar spine↔	EG: PINP↓ S-CTX↓	EG: VAS↓	Li et al. (2014)
Elderly people	2 weeks	Wuqinxi (75) and drug therapy (70)	EG: lumbar spine↑ femoral neck↔ large rotor↔ Ward's triangle↔			Gu and Liu (2021)
Elderly patients with osteoporosis	6 months	Wuqinxi (100) and drug therapy (100)		EG: Ca↔ P↔ BGP↔ ALP↔ PYD↔	EG: VAS↓	Shen et al. (2014)
Osteoporosis	6 months	Yijin Jing (30) and control (30)	EG: lumbar spine↑ femoral neck↑		EG: VAS↓	Jing and Zhang (2008)
Healthy middle-aged females	8 weeks	Yijin Jing (37) and control (34)	EG: BMD↑			Tsai et al. (2008)
Old patients with primary osteoporosis	6 months	Yijin Jing (40) and drug therapy (40)	EG: proximal femur↔		EG: VAS↓ ADL↑	Shen et al. (2018)

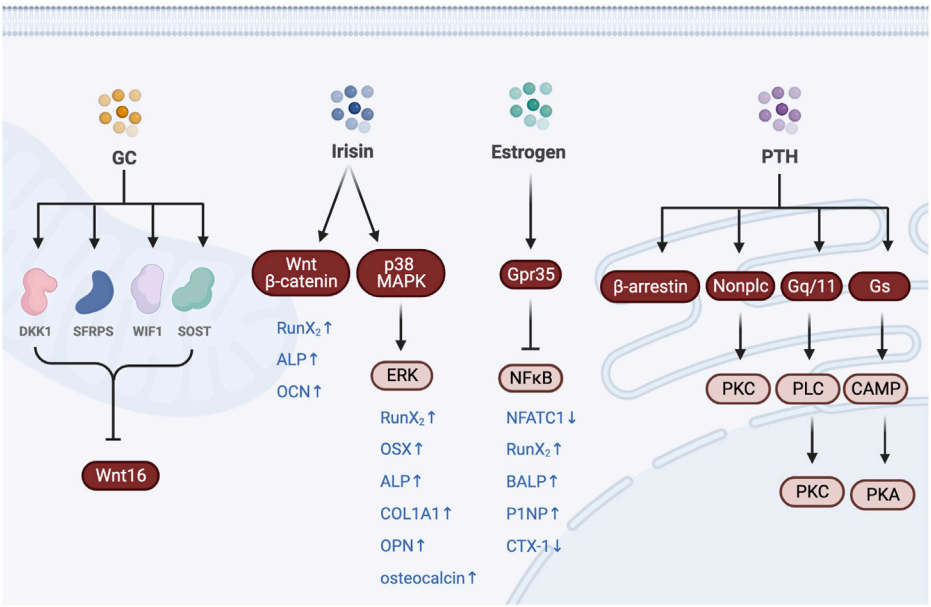

FIGURE 2
Exercise regulates bone metabolism by regulating the hormone secretion of the endocrine system.

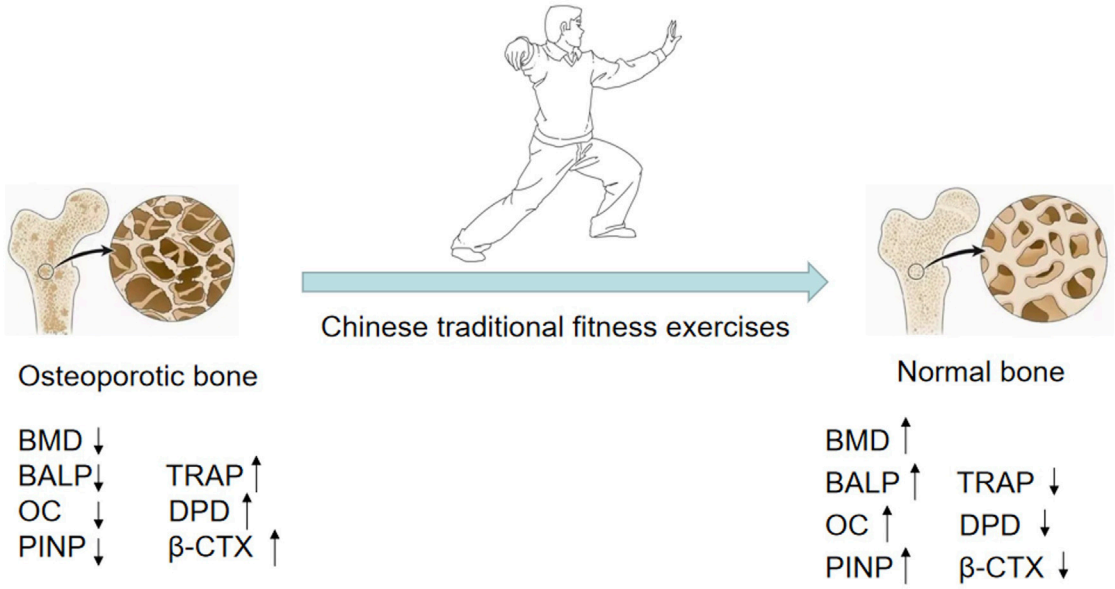

FIGURE 3
Chinese traditional fitness exercises can strengthen the bone, improve bone density, and reduce bone loss.

Chinese traditional fitness exercises are also beneficial to people who are not at risk of osteoporosis. A total of nine articles proved the effects of Taijiquan on the bone health of the elderly, which were detected from the spine, lumbar vertebrae, hip joint, femur, tibia, and ankle joint (Qin et al., 2002; Chan et al., 2004; Qin et al., 2005; Woo et al., 2007; Song, 2008; Zou et al., 2011; Du et al., 2014; Song et al., 2014; Song et al., 2018). Long-term Taijiquan exercise contributed to the increase of the bone mineral density of all parts of the body of the elderly in varying degrees, thereby indicating that Taijiquan played an important role in preventing osteoporosis. Zhou et al. divided 60 postmenopausal women into rope skipping, Mulan boxing, Taijiquan, and Taijiquan groups, and intervened for 10 months. It was concluded that rope skipping, Mulan boxing, and Taijiquan exercises positively affect the bone mineral density of postmenopausal women. Chinese traditional fitness exercises were effective for postmenopausal women to resist bone mass loss and promote the increase of bone mineral density. In addition, the Taijiquan hand-pushing group was comprehensive and effective in increasing the bone mass and bone mineral density (Zhou, 2004). Chen et al. conducted experiments on Baduanjin with healthy middle-aged women, for the women and Baduanjin not only maintained the bone mineral density but also decreased the level of IL-6 in the blood. Thus, it was implicated that Baduanjin effectively prevented common bone loss and was valuable in promoting and maintaining the health status of middle-aged women (Chen et al., 2006). Similarly, Wuqinxi had shown the same effect on maintaining the bone mass and bone mineral density in elderly women (Shen et al., 2013; Gu and Liu, 2021).

Nevertheless, there were certain reservations about the efficacy of Chinese traditional fitness exercises. Hui et al. conducted a randomized controlled trial on a group of middle-aged subjects. After 3 months of Taijiquan intervention, no significant change was found in the bone mineral density before and after the trial (Hui et al., 2015). However, another two studies on Taijiquan have shown that 48 weeks of Taijiquan exercise does not improve bone mineral density in perimenopausal women (Wang et al., 2015; Zhao, 2020). Li et al. believed that the short-term treatment of Wuqinxi for 6 months did not affect the lumbar BMD in patients with primary type I osteoporosis (Li et al., 2014). No significant change was observed in the bone mineral density of the femoral neck and the lumbar vertebrae in postmenopausal women after 24 weeks of a new Wuqinxi exercise but only slightly higher than that in the control group. The data suggested that the new Wuqinxi exercise may play a certain role in delaying bone loss and increasing bone mineral density (Wang et al., 2018). Meanwhile, Tsai et al. concluded that Yijin Jing played an important role in improving muscle endurance and body composition in middle-aged women, but bone strength may not be improved (Tsai et al., 2008). The aforementioned differences may be related to the different movements of traditional sports and different emphases of exercise and may also be related to the age, sex, and initial physical quality of the subjects.

3.2 Chinese traditional fitness exercises improve bone metabolism

Bone metabolism is the basis of ensuring the integrity of the bone composition and structure. The main form of action is the continuous absorption of old bones under the action of osteoclasts, while the continuous synthesis of the new bone is under the action of osteoblasts. A healthy bone is maintained by the dynamic balance between the absorption and reconstruction

of the bone (Ma et al., 2011). Biochemical markers of bone metabolism dynamically reflect the state of bone formation and bone resorption within the organism and also reflect the changes and speed of bone matrix components. Bone alkaline phosphatase (BALP) and alkaline phosphatase (ALP) are related to bone mineralization; N-terminal propeptide (PINP) and the carboxyl-terminal propeptide (PICP) of the procollagen type I are related to a bone matrix, which is proportional to the rate of bone metabolism. In addition, the serum osteocalcin (BGP/OC) is a common biochemical marker of bone formation. The levels of pyridinoline (PYD) and deoxypyridinoline (DPD) in the urine, amino-terminal peptide (NTX), and carboxyl-terminal peptide (CTX) of type I collagen in the blood and serum acid phosphatase (TRAP) are the common markers of bone resorption. Meanwhile, calcium (Ca), phosphorus (P), and the parathyroid hormone (PTH) play regulatory roles in bone resorption and remodeling (Que and Feng, 2014). Biochemical markers of bone metabolism are greatly significant in the early detection, clinical classification, differential diagnosis, treatment monitoring, and screening of osteoporosis (Li, 2007).

Chinese traditional fitness exercises improve the level of bone metabolism in patients with osteoporosis. Wayne et al. found that 9 months of Taijiquan exercise significantly reduced the levels of serum CTX and OC in patients with postmenopausal osteoporosis, thereby suggesting that Taijiquan might reduce bone resorption and improve bone metabolism (Wayne et al., 2012). Similarly, the increase in the levels of serum ALP, Ca, and P and the decrease in the level of DPD/Cr were more effective with a 6-month Baduanjin exercise routine combined with drug intervention compared to simple drug therapy for patients with diabetes mellitus complicated with osteoporosis. In addition, Wuqinxi significantly reduced the levels of serum ALP and BGP in postmenopausal women with osteoporosis and decreased the level of PYD in urine that was equivalent to the effect of a calcium supplement alone (Shen et al., 2014). Li et al. also found that calcium supplements lasting for 9 months combined with modified Wuqinxi were effective in reducing serum PINP and S-CTX levels (Li et al., 2014).

Moreover, conventional Chinese traditional fitness exercise interventions also protected those at risk of osteoporosis. Shen et al. conducted 24 weeks of Taijiquan and resistance training in older people with the highest risk of osteoporosis. The results showed that Taijiquan and resistance training exhibited certain effects against osteoporosis, while the effects were different at different times. The level of the serum BALP in both TC and RT groups increased, but with a 6-week exercise, the level in the TC group was higher than that in the RT group. The serum PTH level in the TC group was significantly higher than that in the RT group with 12 weeks of exercise. The urine Ca^{2+} level in the TC group decreased relatively to the baseline, while that in the RT group increased after 24 weeks of exercise (Shen et al., 2007). It was implicated that the Taijiquan exercise contributed to the regulation of bone metabolism and reducing bone resorption, which showed better effects than resistance exercise. The Taijiquan intervention combined with certain active components in postmenopausal women helped effectively increase the levels of serum BAP and PTH such as green tea polyphenols (Shen et al., 2012).

3.3 Chinese traditional fitness exercises relieve pain

Chinese traditional fitness exercises are simple and slow. The unique movement of "lifting tendons and pulling bones" helps people's muscles, tendons, and bones to be lifted gradually. The movements of pulling, pressing, and stretching are gentler to achieve the purpose of promoting cardiovascular circulation and regulating the metabolism. Chinese traditional fitness exercises play a significant role in relieving the bone, joint, and low back pain caused by bone loss. Meanwhile, for postmenopausal women and elderly patients with osteoporosis, regular Baduanjin exercise also effectively relieved pain. The back pain score of the Baduanjin exercise group was significantly lower than that of the control group and lasted until a subsequent follow-up (Liu et al., 2015). Similar results were also observed for Wuqinxi and Yijin Jing. Both traditional and improved Wuqinxi reduced the pain score of patients with postmenopausal osteoporosis (Jing and Zhang, 2008; Shen et al., 2013; Li et al., 2014; Shen et al., 2014) (Table 1).

Chinese traditional fitness exercises play a significant role in improving and preventing osteoporosis. Nevertheless, there are different types, intensities, and durations of Chinese traditional fitness exercises that may have different effects. Therefore, the exercise should be carried out in combination with the actual situation of various populations. People should carry out the appropriate intensity of pertinent targeted exercises under the guidance of professionals. Generally, the amount of exercise must reach 70%–80% of the maximum heart rate. The more commonly used calculation method is as follows: the optimal heart rate range = (220-age) × (70%–80%). Meanwhile, the exercise frequency is 3–5 times per week, and the time should be controlled between 20 and 60 min each time. It is appropriate not to feel tired the next day. In terms of the duration of exercise, it takes more than 1 year to physiologically increase bone mass. However, strict attention must be paid to safety and preventing falls during the exercise to avoid the occurrence of osteoporotic fractures.

4 Conclusion and prospect

Aerobic exercise is the most appropriate training for people with osteoporosis because of their fragile bones. As one of the manifestations of aerobic exercise, Chinese traditional fitness exercises have significant advantages in preventing and treating osteoporosis. In addition, Chinese traditional fitness exercises increase bone mass, decelerate bone loss, and improve bone metabolism, and play a significant role in improving balance, reducing fall risks, relieving pain, and improving the quality of life. Chinese traditional fitness sports need low skill requirements and have a higher safety index than ordinary aerobic exercises. It is rich in various movements, which meet the different exercise needs of different people. Chinese traditional fitness sports bring enormous interest and greatly improve patients' compliance. However, there are certain limitations to Chinese traditional fitness exercises. The effects are generally slow, while the courses of treatment are relatively long. In addition, there is an upper limit for the improvement of bone mass. When the bone mass increases

up to a certain level, continuous exercise may still slow down the rate of bone loss, but it will not continue to increase bone mineral density (Benedetti et al., 2018). However, certain skills are necessary for some of the movements of Chinese traditional fitness exercises. If unqualified or wrong exercises are carried out for a long time, they may reduce the curative effects and even cause injury.

The current studies mostly focused on the clinical effects such as the bone mineral density, bone metabolism index, and pain, but there was no way to verify them by special animal experiments. There is a lack of discussion of the mechanism at the present stage. In addition, the sample size of most clinical trials was relatively small, and some trials did not strictly follow the random and double-blinded principles. Thus, the problem of the short intervention time and lack of long-term follow-up reports remains. Moreover, there is no set of unified guidelines on how to choose the appropriate exercise therapy, and there is no unified standard for the time and frequency of each exercise. Therefore, it is essential to formulate a standardized, reasonable, and scientific exercise prescription, and thus, this will be our next research direction.

Our work was performed concerning the effects of Chinese traditional fitness exercises against osteoporosis. This review may contribute to the development of more targeted methods to improve and prevent osteoporosis to provide theoretical references. Herein, molecular mechanisms of exercises in improving osteoporosis are also summarized, which can further aid the development of rational therapies based on Chinese traditional fitness exercises. We believe that this review may lay the foundation for the development of appropriate treatments to prevent and treat osteoporosis.

Author contributions

ZW and X-AZ: conceptualization, project administration, and funding acquisition. WS, ZW, and X-AZ: writing, review, and editing. All authors contributed to the article and approved the submitted version.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Automated recognition of asymmetric gait and fatigue gait using ground reaction force data

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Introduction: The purpose of this study was to evaluate the effect of running-induced fatigue on the characteristic asymmetry of running gait and to identify non-linear differences in bilateral lower limbs and fatigued gait by building a machine learning model.

Methods: Data on bilateral lower limb three-dimensional ground reaction forces were collected from 14 male amateur runners before and after a running-induced fatigue experiment. The symmetry function (SF) was used to assess the degree of symmetry of running gait. Statistical parameter mapping (Paired sample *T*-test) algorithm was used to examine bilateral lower limb differences and asymmetry changes pre- and post-fatigue of time series data. The support vector ma-chine (SVM) algorithm was used to recognize the gait characteristics of both lower limbs before and after fatigue and to build the optimal algorithm model by setting different kernel functions.

Results: The results showed that the ground reaction forces were asymmetrical (SF > 0.5) both pre-and post-fatigue and mainly concentrated in the medial-lateral direction. The asymmetry of the medial-lateral direction increased significantly after fatigue ($p < 0.05$). In addition, we concluded that the polynomial kernel function could make the SVM model the most accurate in classifying left and right gait features (accuracy of 85.3%, 82.4%, and 82.4% in medial-lateral, anterior-posterior and vertical directions, respectively). Gaussian radial basis kernel function was the optimal kernel function of the SVM algorithm model for fatigue gait recognition in the medial-lateral and vertical directions (accuracy of 54.2% and 62.5%, respectively). Moreover, polynomial was the optimal kernel function of the anterior-posterior di-rection (accuracy = 54.2%).

Discussion: We proved in this study that the SVM algorithm model depicted good performance in identifying asymmetric and fatigue gaits. These findings can provide implications for running injury prevention, movement monitoring, and gait assessment.

KEYWORDS

symmetry function, fatigue gait, statistical parameter mapping, running, support vector machine

1 Introduction

Running biomechanics has received widespread attention from runners, coaches and researchers for the past 30 years (Furlong and Egginton, 2018; Gao, 2022; Xu et al., 2022). Typically, the mean value of bilateral variables (Winter, 1984; Grabowski and Kram, 2008) or the default complete symmetry of both limbs (Kyröläinen et al., 2005; Gao et al., 2020c) was widely used in running biomechanical research. Bilateral limb asymmetry is not considered in these studies since complete symmetry of gait is assumed. Although these methods mentioned above can describe the motion well, it also ignores the false claims and misleading interpretations caused by the asymmetry of bilateral variables (Beck et al., 2018). Neuromuscular asymmetry is a widespread phenomenon occurred in functional tasks (Radzak et al., 2017). Few previous studies have considered that the biomechanical asymmetry of running gait (Beck et al., 2018; Gao et al., 2020b; Mastalerz, 2021), even though this phenomenon is common among healthy people (Furlong and Egginton, 2018; Gao et al., 2020c). Quantitative gait characteristics (e.g., parameters of time and parameters of space) and qualitative gait characteristics (e.g., gait variability and gait asymmetry) are related to running-related injuries, especially among amateur runners (Nakayama et al., 2010; García-Pinillos et al., 2020). Similarly, the effects of gait asymmetry are also an important consideration for motor performance (Simoni et al., 2021). Previous studies have shown that a 10% increase in asymmetry of vertical ground reaction force (GRF) leads to a 3.5% increase in net metabolic power during running (Beck et al., 2018). Another finding was that increased foot contact time asymmetry was associated with increased metabolic costs of running (Beck et al., 2018). Zifchock et al. (2006) found that the asymmetry was 49.8% and 37.5% at the running speed of 3.65 m/s by evaluating the peak lateral and medial GRF of bilateral lower limbs in the running process of healthy individuals, similarly by Williams et al. (1987) reported that 13.8% and 20.2% asymmetry of peak lateral and medial GRF at the speed of 5.36 m/s, suggesting that greater symmetry is associated with faster running speeds. The same conclusion regarding walking gait was reported by previous study (Goble et al., 2003).

Bilateral limb asymmetry may not be evident during the initial stages of running but may arise as muscle fatigue and/or exercise intensity changes (Arampatzis et al., 1999; Xiang et al., 2022a). Mizrahi et al. (2000) demonstrated that running-induced lower limb muscle fatigue increases the tibia's vertical acceleration during heel strike, suggesting that more load accumulates in the shank after running fatigued. Research by Mastalerz (2021) reported that the asymmetry of muscle activity on bilateral limbs was increased with the occurrence of running fatigue, suggesting that bilateral lower extremity muscles have different fatigue resistance. Typically, fatigue in the lower limb muscles reduces the cushioning effect of the muscles on the joint load. Therefore, the joint load tends to focus on one limb during running-induced fatigue, and the risk of unilateral lower limb injury increases (Furlong and Egginton, 2018). Radzak and colleagues (Radzak et al., 2017) found an increase in knee stiffness and internal rotation angle asymmetry after performing a running-induced fatigue test. In addition, a recent systematic review of the relationship between fatigue and symmetry shows that the effect of running fatigue on bilateral lower limb symmetry is not apparent, and the reasons may be related to experimental design, quantitative indicators, and test protocol (Heil et al., 2020). Running-induced fatigue may cause subtle or worsening existing gait asymmetry (Gao et al., 2020a). The assessment of bilateral limb asymmetry in functional movement is an essential measure of injury prediction and screening (Kiesel et al., 2007). In addition, a recent study demonstrates that reduced asymmetry in vertical impact peaks was associated with running injuries (Ceyssens et al., 2019). Therefore, a further factor that contributes to damage during long-distant running tasks is the presence of asymmetry (Gao et al., 2022b). It is particularly important to properly evaluate the effects exercise-induced fatigue on interlimb asymmetry to understand the underlying mechanisms of non-contact injury and improve exercise efficiency (Smeets et al., 2019).

The evaluation approaches of running gait symmetry have been widely utilized in biomechanical studies in past decades (Zifchock et al., 2006; Nigg et al., 2013; Radzak et al., 2017). Typically, the traditional classification theory of symmetry, such as symmetry index (SI) (Herzog et al., 1989) and symmetry angle (SA), was based on discrete variables like the mean peak GRF (Viteckova et al., 2018). However, the time-series information may be lost in the evaluation process of continuous data, such as a complete period of gait stance (Tabor et al., 2021). Therefore, symmetry function (SF), a novel symmetry checking method, was proposed by Nigg to evaluate continuous data with time dimensions (Nigg et al., 2013). Moreover, linear principal component analysis (PCA) has been used in the previous study to extract the gait characteristics of bilateral lower limbs and examine the degree of gait symmetry (Sadeghi, 2003). However, the limitation of SF and PCA is that only linear features of gait can be extracted, which reduces the model's sensitivity in evaluating the non-linear features of time series parameters (Chau, 2001). The previous study demonstrated that the support vector machine (SVM) has a high generalization ability for dichotomous data (Figueiredo et al., 2018). The optimal separating hyperplane was created by maximizes the distance of separation in SVM model (Figueiredo et al., 2018). Moreover, it can transform the matrix into a higher dimensional space for classification by setting the different types of kernel functions (Figueiredo et al., 2018; Zhang et al., 2023). SVM algorithms was widely used in gait patterns recognition, such as differences of young and elderly populations (Eskofier et al., 2013), competitive and recreational runners (Clermont et al., 2017), barefoot and shod populations (Gao et al., 2022a). Given the gait differences caused by running fatigue, there is still a lack of understanding of the gait characteristics of dominant and non-dominant limbs and how these differences are attributed to limb injury prevention and function performance. Furthermore, the automatic detection of asymmetric gait and fatigue gait was lacking in the previous studies. Consequently, the SVM based on the structural risk minimization principle, were adopted to evaluate bilateral lower limb non-linear differences in this study (Chau, 2001).

This study aims to examine the asymmetry changes of bilateral gait characteristics before and after running fatigue and to develop a SVM machine learning model for asymmetry and fatigue gait recognition. It was hypothesized that the GRFs on both sides are asymmetrical, increasing with fatigue intervention. The second hypothesis is that non-linear kernels perform better than linear kernels for SVM models that recognize asymmetric and fatigue gaits.

TABLE 1 Descriptive characteristics of 14 participants.

Information	Mean	SD
Age (year)	22.93	1.07
Height (cm)	176.83	2.61
Weight (kg)	70.17	6.57
BMI (kg/m^2)	23.19	0.96

2 Materials and methods

2.1 Participants

This study recruited fourteen male amateur runners who ran at least twice a week for less than 45 min or less than 10 km. The demographic information is given in Table 1. All subjects were asked to complete a questionnaire to determine whether the right limb was the dominant limb (the ball kicking leg) and had not suffered any lower limb or pelvic injuries in the previous 6 months. In addition, individuals with a history of lower extremity surgery were also excluded. The subjects have agreed in writing all the experiment contents, and the Ethics Committee of Ningbo University has approved this test protocol.

2.2 Data collection

Before this experiment, subjects were fully familiarized with the environment and processed a 10-min treadmill warm-up. The complete experiment was divided into three parts: data collection of GRFs of bilateral lower limbs before running fatigue, running-induced fatigue experiment (on the treadmill), and collection of the GRFs of bilateral lower limbs after running fatigue, as shown in Figure 1. The same acquisition process was applied with two GRF data collection experiences (pre- and post-fatigued).

As shown in Figure 1A1, a 3-dimensions force measuring plate (Sampling rate: 1,000 Hz) was embedded on a 20-m runway (Kistler, Winterthur, Switzerland). The force measuring plate was calibrated before each subject performed the test task. The participants determine the optimal distance from the force plate by familiarizing runway to increase the likelihood of effective contact between bilateral feet and the force plate during the runway trials. Whereafter, participants were instructed to run across a runway at a comfortable speed after familiarization and practices on the runway. During testing, the right foot didn't land in the middle of the plate and the participants has active aiming force plate behavior were excluded (Figure 1A1). After a successful running trial has been performed, the same running mode was replaced with the left foot. until the three successful GRF data of the left and right sides were respectively collected (Figure 1A2). The second step was to implement the Running-induced Fatigue Protocol Test developed by Koblbauer and their colleagues (Koblbauer et al., 2014).

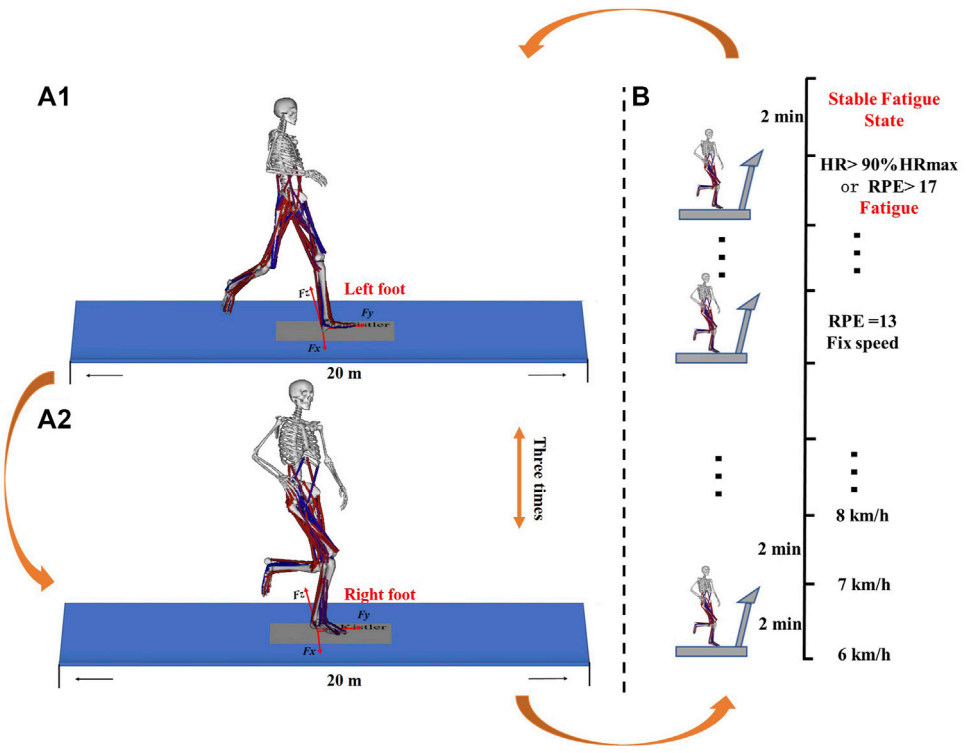

FIGURE 1
Bilateral data collection execution process. **(A1)** Right GRF data collection process. **(A2)** Left GRF data collection process. **(B)** Running-induced Fatigue Protocol Test implementation process.

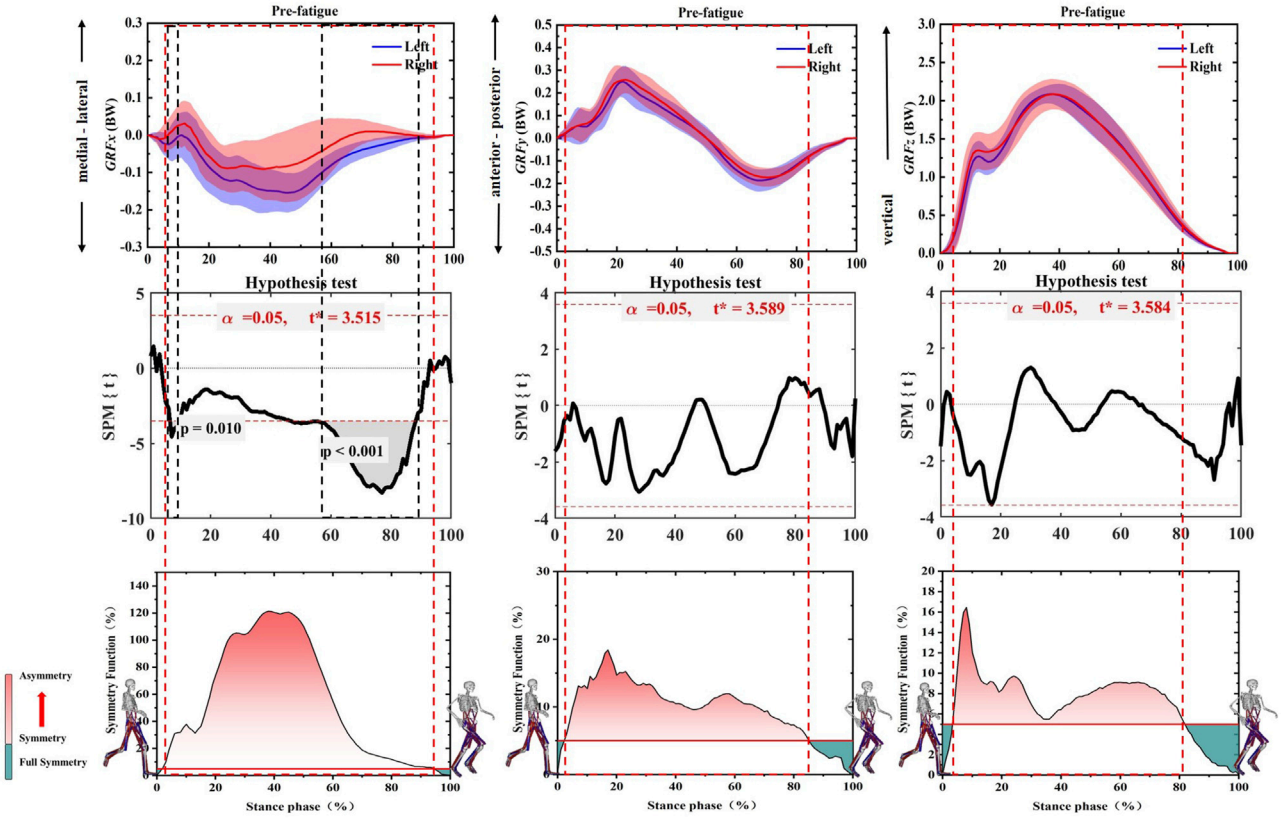

FIGURE 2
Illustration of SF degree and SPM test of GRFx, GRFy, and GRFz during the whole gait stance phase before Running-induced Fatigue Protocol Test. Note: The red dotted boxes represent asymmetrical degrees. The black dotted boxes represent a significant difference ($p < 0.05$). The red fill represents the degree of asymmetry, and the darker the color, the more asymmetry. The green fill represents full symmetry. GRFx: medial-lateral direction of GRF, GRFy: anterior-posterior direction of GRF, GRFz: vertical direction of GRF.

The specific execution procedures have been reported in the previous study (Koblbauer et al., 2014). As shown in Figure 1A2, Heart rate (Polar RS100, Polar Electro Oy, Woodbury, NY, USA) and RPE (Ratings of Perceived Exertion, RPE, 6–20 Borg's scale) were collected every 2 min throughout the procedure. The running speed increases by 1 km/h every 2 min until RPE = 13. Continue running at a fixed speed for 2 min after fatigue (HR>90%HRmax or RPE>17, HRmax = 200-age) (Gao et al., 2020a). The GRF data collection process was repeated until three successful data were collected on the left and right sides after the running-induced fatigue protocol test (Figure 1A1).

2.3 Data processing

GRF time series data of one support period were determined by vertical GRF = 30 N as the heel initial contact and toe-off phases (Quan et al., 2021). The GRF data of the time series were normalized to 101 frames by Cubic B-spline interpolation or extrapolation function in MATLAB (Version: R2019a, The MathWorks, Natick, MA, United States). All GRF data were standardized by individual body weight (GRF (BW) is equal to GRF divided by 10 times the body weight of the individual).

SF developed by Nigg and colleagues (Nigg et al., 2013) has high robustness for quantifying the symmetry of time series data, which can effectively avoid the data loss caused by the symmetry evaluation of discrete data. The symmetry time-varying information of 101-time points in SF can reflect the supporting stage of gait (Eq. 1). The degree of asymmetry is positively correlated with the value of SF, where the closer SF is to 0, the more symmetric the parameters are (Eq. 2). ±5% was set as an asymmetric threshold in the evaluation of this study (Winiarski et al., 2021).

$$\mathbf{SF} = \int_{t=t_1}^{t_2} \mathbf{A}|\mathbf{x_r}(\mathbf{t}) - \mathbf{x_l}(\mathbf{t})|\mathbf{dt} \tag{1}$$

$$\mathbf{A} = \frac{2}{\mathbf{range}(\mathbf{x_r}(\mathbf{t})) + \mathbf{range}(\mathbf{x_l}(\mathbf{t}))} \tag{2}$$

Where $\mathbf{x_r}(\mathbf{t})$ and $\mathbf{x_l}(\mathbf{t})$ were defined as the value of the GRF for the right and left foot at the time t, respectively. The $\mathbf{t_1}$ is the time at heel strike point during stance phase and $\mathbf{t_2}$ is the time at take-off point during the stance phase.

2.4 Machine learning model building

SVM is a machine learning algorithm that realizes data dichotomy through case learning (Teufl et al., 2019). It has been widely used in biology and medicine (Noble, 2006). Such as the automatic classification of DNA sequences, gene

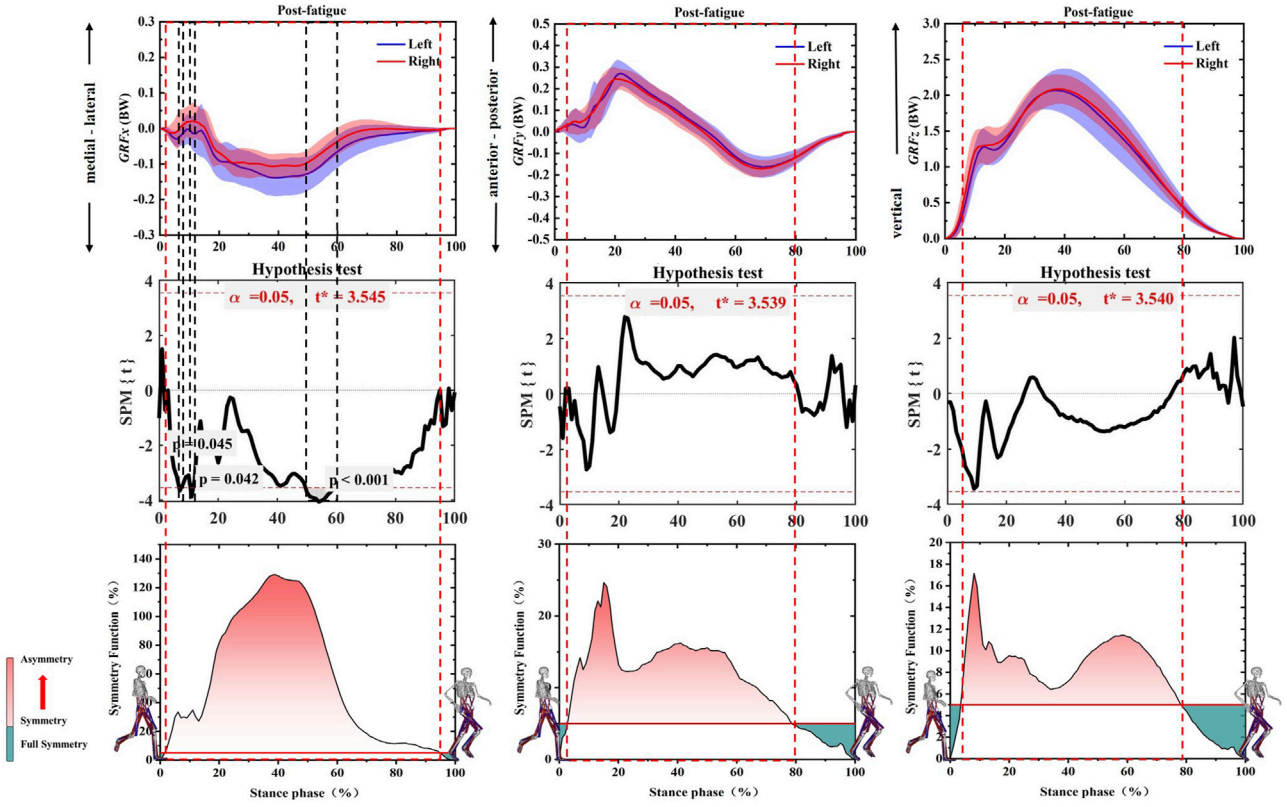

FIGURE 3
Illustration of SF degree and SPM test of GRFx, GRFy, and GRFz during the whole gait stance phase after Running-induced Fatigue Protocol Test. Note: The red dotted boxes represent asymmetrical intervals. The black dotted boxes represent a Significant difference. The red fill represents the degree of asymmetry, and the darker the color, the more asymmetry. The green fill represents full symmetry. GRFx: medial-lateral direction of GRF, GRFy: anterior-posterior direction of GRF, GRFz: vertical direction of GRF.

expression profiles from tumor samples or peripheral fluids, microarray expression profiles, and mass spectrometry (Golub et al., 1999). Compared with other machine learning algorithms, such as neural networks, the SVM algorithm based on supervised learning has better robustness and generalization for gait symmetry analysis (Wu and Wu, 2015). In this study, six SVM models were constructed to evaluate the characteristic differences of GRF in the medial-lateral (X), anterior-posterior (Y) and vertical (Z) directions of left and right feet and before and after fatigue. Among them, the left GRF and SF of Pre-fatigue belong to feature 1 (X, Y, and Z directions, respectively). The right GRF and SF of Post-fatigued belong to feature 2 (X, Y, and Z directions, respectively). Optimal symmetric gait and fatigue gait recognition model by adjusting the non-linear kernel function. The kernel function $K(x_i, x) = \phi(x_i)\phi(x)$ based on Mercer theorem (Vapnik, 1999).

The difference of non-linear gait features between dominant and non-dominant limbs was identified by the SVM algorithm in MATLAB (Version: R2019a, The MathWorks, Natick, MA, United States). The first step is to map the non-linear function ϕ: $R^b \rightarrow F$ to high dimensional feature space W. In addition, the optimal linear classification plane f in W was determined by feedback learning. The training set data sample D is assumed by Eq. 3.

$$\mathbf{D} = \{(\mathbf{x_i}, \mathbf{y_i})\}_{\mathbf{i=1}}^{\mathbf{n}} \tag{3}$$

Where each raw output quantity is defined as $x_i \in R^b$ B and n represent the original input data's feature dimension and sample size, respectively.

$$\mathbf{f}: \mathbf{R^b} \rightarrow \{-\mathbf{1}, +\mathbf{1}\} \tag{4}$$

$$\mathbf{x_i} \rightarrow \mathbf{y_i} \tag{5}$$

In the training set data, two gait pattern data are mapped to the corresponding class flag space $y_i \in \{-1, +1\}$, − 1 in each SVM model in this study, the parameters in features 1 and + 1 are the parameters in feature 2. In addition, the optimal linear classification surface f in the model is obtained by Eq. 6

$$\mathbf{f(x)} = \mathbf{sgn}\left(\sum_{\mathbf{i=1}}^{\mathbf{n}} \boldsymbol{\beta}_\mathbf{i}\mathbf{y_i}\mathbf{K(x_i, x)} + \mathbf{b}\right) \tag{6}$$

$$\mathbf{minW}(\boldsymbol{\beta}) = -\boldsymbol{\beta}^\mathbf{T}\mathbf{I} + \frac{\mathbf{1}}{\mathbf{2}}\boldsymbol{\beta}^\mathbf{T}\mathbf{D}\boldsymbol{\beta} \tag{7}$$

$$\mathbf{s.t.}\ \boldsymbol{\beta}^\mathbf{T}\mathbf{y} = \mathbf{0}; \boldsymbol{\beta}_\mathbf{i} \in [\mathbf{0}, \mathbf{C}] \tag{8}$$

where b and β_i are defined as estimation bias and optimal classification hyperplane coefficients for training set data, respectively. C is the penalty factor, which determines the distance between the support vector and the decision plane. Parameter G is mainly used for height mapping of low-

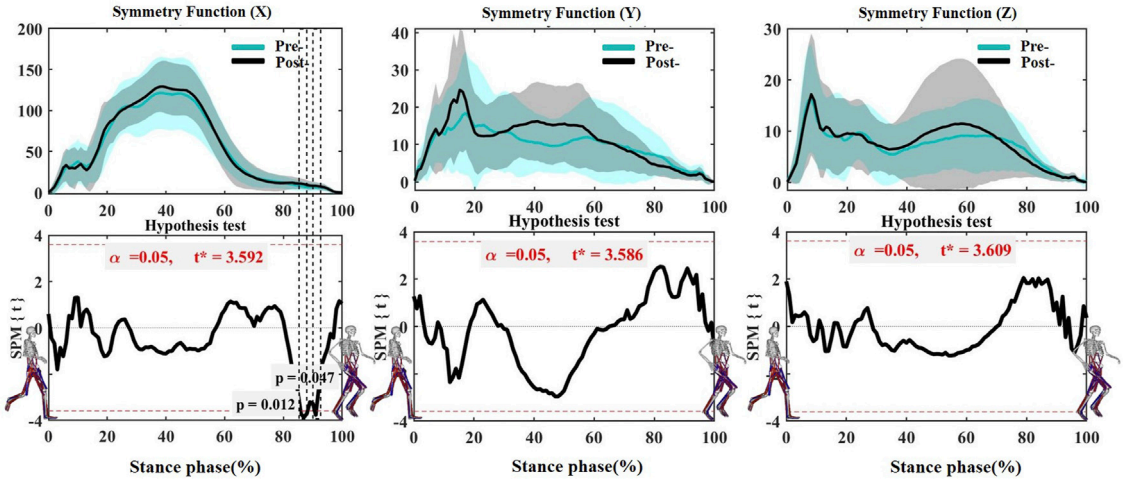

FIGURE 4
Comparing the mean values of SF of GRFx, GRFy, and GRFz from all participants between Pre-fatigue and Post-fatigue. Note: Pre-is Pre-fatigue, and Post-is Post-fatigue. The black dotted boxes represent a significant difference ($p < 0.05$). GRFx: medial-lateral direction of GRF, GRFy: anterior-posterior direction of GRF, GRFz: vertical direction of GRF.

dimensional samples, and the larger the value of g, the higher the dimension of the mapping. Optimal classification hyperplane coefficients can be obtained by solving quadratic programming problems in Eqs 7, 8. In the SVM model with stable inseparable linear data, the balance between the maximum interval and the minimum training error is adjusted by setting the best *C* and best G parameter.

In this study, the accuracy of the SVM model was evaluated by the five-fold cross-validation method (Begg and Kamruzzaman, 2005). The optimal machine learning model was determined based on different kernel function types and parameters by adjustment. The following three commonly used kernel function types were selected to evaluate the generalization ability of the SVM model in this study.

Linear kernel function (LINEAR) is shown in Eq. 9:

$$\mathbf{K}(\mathbf{x_i}, \mathbf{x}) = \mathbf{x_i x} \tag{9}$$

Gaussian radial basis kernel function (RBF) is shown in Eq. 10:

$$\mathbf{K}(\mathbf{x_i}, \mathbf{x}) \exp\left(-\frac{\| \mathbf{x_i} - \mathbf{x} \|^2}{2\sigma^2}\right) \tag{10}$$

The polynomial kernel function (POLY) is shown in Eq. 11:

$$\mathbf{K}(\mathbf{x_i}, \mathbf{x}) = (\mathbf{x_i x} + \mathbf{1})^{\mathbf{d}} \tag{11}$$

where the *d* is defined as the order of the POLY.

Classification Accuracy (ACC), Sensitivity (SEN), and Specificity percentage (SEP) are used to evaluate the performance of classification machine learning models (Begg and Kamruzzaman, 2005). Thereinto, *Ac* (12) was used to evaluate the model's ability to strive for recognition of the two features. *Se* (13) and *Sp* (14) were used to evaluate the model's ability to correctly identify the first feature and the second feature, respectively.

$$\mathbf{ACC} = \frac{\mathbf{T_1} + \mathbf{T_2}}{\mathbf{T_1} + \mathbf{F_1} + \mathbf{T_2} + \mathbf{F_2}} \times \mathbf{100\%} \tag{12}$$

$$\mathbf{SEN} = \frac{\mathbf{T_1}}{\mathbf{T_1} + \mathbf{F_2}} \times \mathbf{100\%} \tag{13}$$

$$\mathbf{SEP} = \frac{\mathbf{T_2}}{\mathbf{T_2} + \mathbf{F_1}} \times \mathbf{100\%} \tag{14}$$

where, T1 and T2 are the correct recognition formats of feature 1 and feature 2 by the SVM model, and F1 and F2 are the number of feature 1 and feature 2 incorrectly recognized by the SVM model, respectively.

2.5 Statistical analysis

In this study, The Shapiro-Wilks test was used to verify the normality of the data prior to statistical analysis. SF was used to evaluate the symmetry degree of bilateral GRF, and the paired sample *T*-test in the statistical Parameter Mapping (SPM) algorithm was used as the different test of bilateral gait characteristics. In addition, the paired sample *T*-test in SPM was also used to check SF of GRF changes in X, Y, and Z directions before and after fatigue in MATLAB (Version: R2019a, The MathWorks, Natick, MA, United States). The significance level was set to 0.05.

3 Results

3.1 Pre-fatigue biomechanical variable

There are asymmetries in all three directions before Running-induced Fatigue Protocol Test, as shown in Figure 2. Specifically, the SF of GRFx rise steadily until the peak value of 1.2 at 3%–40% of the

TABLE 2 Comparison of results from the different SVM classification algorithms designed for SVM models of GRFx, GRFy, and GRFz. Note: The best kernel type is highlighted in bold. GRFx: medial-lateral direction of GRF, GRFy: anterior-posterior direction of GRF, GRFz: vertical direction of GRF.

	Kernel	Cross-validation results			Model precision		
		Best C	Best G	CVAcc	ACC	SEN	SEP
	LINEAR	35.506	0.036	0.836	0.794	0.857	0.75
GRF*x*	RBF	35.506	0.036	0.836	0.824	0.762	0.923
	POLY	35.506	0.036	0.836	**0.853**	**0.842**	**0.933**
	LINEAR	3.605	0.113	0.821	0.794	0.813	0.778
GRF*y*	RBF	3.605	0.113	0.821	0.676	0.667	0.688
	POLY	3.605	0.113	0.821	**0.824**	**0.762**	**0.923**
	LINEAR	14.929	0.069	0.784	0.677	0.75	0.652
GRF*z*	RBF	14.929	0.069	0.784	0.824	0.824	0.824
	POLY	14.929	0.069	0.784	**0.824**	**0.867**	**0.789**

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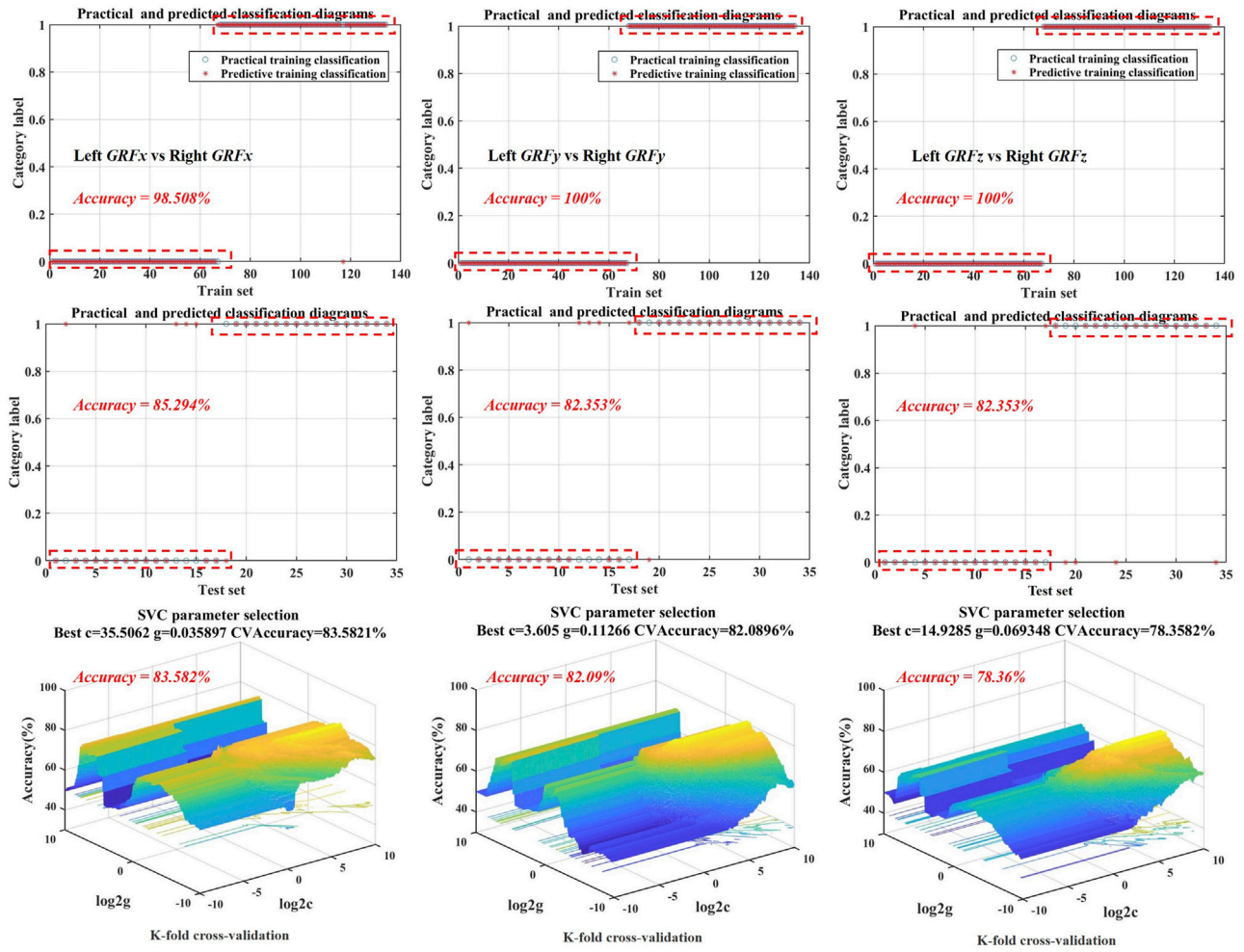

FIGURE 5
The training set, test set, and cross-validation model accuracy visualization are based on optimal kernel function selection for SVM models of GRFx, GRFy, and GRFz. GRFx: medial-lateral direction of GRF, GRFy: anterior-posterior direction of GRF, GRFz: vertical direction of GRF.

gait stance phase, and then it declines to 0.05 at 95% of the gait stance phase. In addition, the SPM results showed that the non-dominant limb appears to have greater medial direction force on 7%–10% ($p = 0.010$, t = 3.515) and 58%–90% of the gait stance phase ($p < 0.001$, t = 3.515). In addition, the SF of GRFy and GRFz were more than 0.05 in 3%–85% and 4%–80% of the gait stance phase, respectively.

3.2 Post-fatigued biomechanical variables

As shown in Figure 3, asymmetry of GRFx, GRFy, and GRFz was observed on both sides after the Running-induced Fatigue Protocol Test. The SF of GRFx exceeded 1.0 in 30%–55% of the gait stance phase. the SPM results showed that the left GRFx exhibited greater medial force on 8%–9% ($p = 0.045$, $t = 3.545$), 10%–12% ($p = 0.042$, $t = 3.545$), and 50%–60% ($p < 0.001$, $t = 3.545$) of the gait stance phase. Moreover, the SF of GRFy and GRFz were more than 0.05 in 3%–80% and 5%–79% of the gait stance phase, respectively.

3.3 Symmetry function of pre- and post-fatigue

According to the SPM inspection results (Figure 4), Only SF of GRFx changed significantly after the Running-induced Fatigue Protocol Test. SF of post-fatigued was significantly higher than that pre-fatigue at 86%–88% ($p = 0.012$, $t = 3.592$) and 90%–92% ($p = 0.047$, $t = 3.592$) of the gait stance phase. GRFy and GRFz did not change significantly after the Running-induced Fatigue Protocol Test.

3.4 SVM model generalization ability

3.4.1 Left and right SVM models

It can be seen from Table 2 that the optimal kernel functions of SVM models of GRFx, GRFy, and GRFz are all POLY. ACC, SEN, and SEP of the SVM model of GRFx were achieved at 85.3%, 84.2%, and 93.3%, respectively. Through model cross-validation, best C, Best G, and CVAcc are 33.506, 0.036, and 0.836, respectively. The

TABLE 3 Comparison of results from the different SVM classification algorithms designed for SVM models of SFx, SFy, and SFz before and after the Running-induced Fatigue Protocol Test. Note: The best kernel type is highlighted in bold, SFx: medial-lateral direction of SF, SFy: anterior-posterior direction of SF, SFz: vertical direction of SF.

	Kernel	Cross-validation results			Model precision		
		Best C	Best G	CVAcc	ACC	SEN	SEP
	LINEAR	1.569	0.043	0.75	0.5	0.5	0.5
SF*x*	**RBF**	1.569	0.043	0.75	**0.625**	**0.667**	**0.6**
	POLY	1.569	0.043	0.75	0.625	0.636	0.615
	LINEAR	20.393	0.002	0.633	0.542	0.556	0.533
SF*y*	RBF	20.393	0.002	0.633	0.56	0.6	0.533
	POLY	20.393	0.002	0.633	**0.542**	**1**	**0.522**
	LINEAR	0.001	9.514	0.617	0.5	0	0.5
SFz	**RBF**	0.001	9.514	0.617	**0.625**	**0.75**	**0.63**
	POLY	0.001	9.514	0.617	0.5	0.5	0.5

prediction accuracy of the test set of GRFx reached 85.294%, as shown in Table 2 and Figure 5. In addition, ACC, SEN, and SEP of GRFy SVM model were achieved at 82.4%, 76.2%, and 92.3%, respectively. The prediction accuracy of the test set of GRFy reached 82.35%, as shown in Figure 4. Through model cross-validation of the SVM model of GRFy, the Best C, Best G, and CVAcc are 3.605, 0.113, and 0.821, respectively. Moreover, the SVM Model of GRFz showed 82.35% prediction accuracy for the test set (Figure 4). ACC, SEN, and SEP of the SVM model of GRFz were achieved at 82.4%, 86.7%, and 78.9%, respectively, and the Best C, Best G, and CVAcc are 14.929, 0.069, and 0.784, respectively.

3.4.2 SF of pre-fatigue and post-fatigued SVM models

It can be seen from Table 3 that the optimal kernel functions of SVM models of SFx, SFz are POLY, and the optimal kernel function of SFy is POLY. ACC, SEN, and SEP of the SVM model of SFx were achieved at 62.5%, 66.7%, and 60%, respectively. Through model cross-validation, Best C, Best G, and CVAcc are 1.569, 0.043, and 0.75, respectively. The prediction accuracy of test set reached 62.5%, as shown in Table 3 and Figure 5. In addition, ACC, SEN, and SEP SVM models of SFy were achieved 54.2%, 100%, and 52.2%, respectively. In addition, the prediction accuracy of the test set reached 54.167%, as shown in Figure 5. Through model cross-validation of the SVM model of SFy, the Best C, Best G, and CVAcc are 20.393, 0.002, and 0.633, respectively. Moreover, the SVM model of SFz showed 62.5% prediction accuracy of the test set (Figure 4). ACC, SEN, and SEP of SVM model of SFz were achieved 62.5%, 74%, and 53%, respectively, and the Best C, Best G, and CVAcc are 0.001, 9.514, and 0.617, respectively.

4 Discussion

This study aimed to examine the effect of running fatigue on bilateral GRF symmetry and develop SVM models to realize asymmetric gait and fatigue gait automated recognition. We found that GRF of bilateral lower limbs was asymmetric in all three directions (medial-lateral, anterior-posterior and vertical) at pre-fatigue states, especially a significantly worsened of asymmetry with the occurrence of running fatigue in the medial-lateral direction. In addition, the SVM model with POLY kernel has demonstrated to have higher accuracy for feature extraction of symmetric gait, while the SVM model with RBF kernel has higher accuracy for fatigue gait automated recognition in anterior-posterior and vertical directions. Furthermore, the POLY kernel had highlighted to have relatively higher accuracy for fatigue gait recognition in the vertical direction. In general, the results of this study are consistent with the previous hypotheses.

Gait asymmetry in healthy individuals may be related to the functional attributes of bilateral limbs (Sadeghi et al., 2000; Pan et al., 2023). Previous study have emphasized that the dominant limb usually plays a gait propulsion role while opposing limbs contribute to gait support and control (Sadeghi et al., 2000). The dominant limb during the gait cycle was associated with more power generation, which was mainly reflected in the positive anterior-posterior GRF impulse (Seeley et al., 2008). Therefore, the non-dominant limb is subjected to significant negative anterior-posterior GRF impulse (Potdevin et al., 2008). Interestingly, this study found statistical differences in medial-lateral direction of GRF during the push-off phase before the fatigue intervention, suggesting the neuromuscular control asymmetry in healthy individuals (Radzak et al., 2017). The non-dominant limb maintains the stability of the gait stance phase through the greater medial force (Sadeghi et al., 2000). Similarly, the larger SF of medial-lateral direction of GRF present in the stance phase of the entire stance phase can also explain this idea. The only slight asymmetry of vertical direction of GRF (SF < 18) and no significant difference before and after fatigue were reported in the current study, which contradicted the previous reports that vertical direction of GRF was a major variable of symmetry in running gait (Gao et al., 2020b).

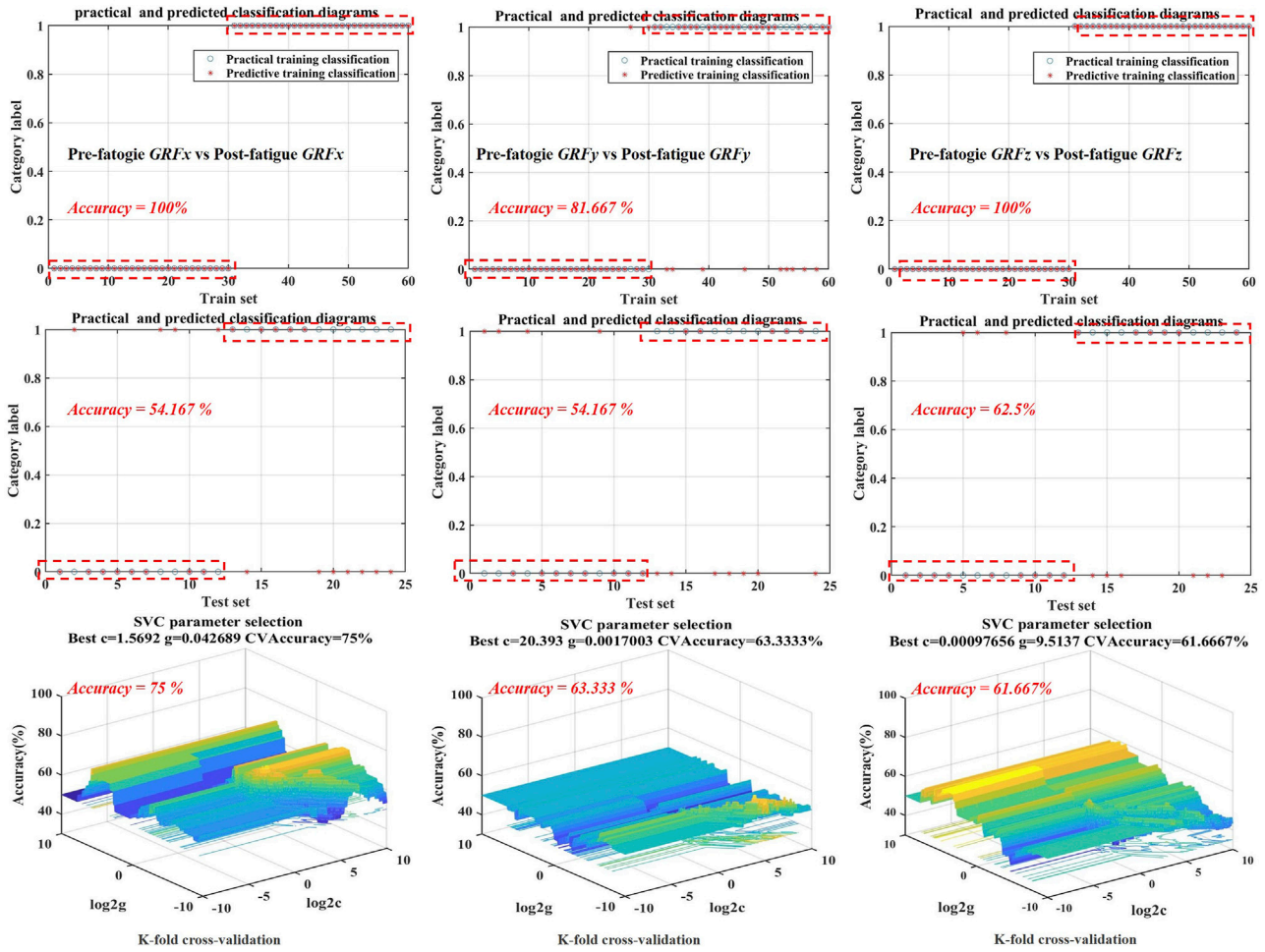

FIGURE 6
Training set, test set and cross-validation model accuracy visualization are based on optimal kernel function selection for SVM models of SF for SVM models of SF*x*, SF*y* and SF*z* before and after Running-induced Fatigue Protocol Test. SFx: medial-lateral direction of SF, SFy: anterior-posterior direction of SF, SFz: vertical direction of SF.

The kinematics and kinetics variables of bilateral lower limbs may be changed due to the weakened central nervous control over muscles during long-distance running (Quan et al., 2021). Our findings are in line with a previous argument that gait asymmetry increases with fatigue (Gao et al., 2020a). The current study report that the asymmetry was mainly found in the medial-lateral direction, and it was observed to occur in the heel contact stage (8%–9% and 10%–12%) and the mid-foot forward transition stage (50%–60%), suggesting that the more significant medial load exist in the non-dominant foot during running gait after fatigue intervention. This phenomenon may be the potential cause of ankle pronation and arch collapse after long-distance running (Fourchet et al., 2015). Moreover, The asymmetry of anterior-posterior and vertical direction of GRF was observed to occur before the push-off period, which was consistent with the performance before fatigue, suggesting that fatigue did not affect the asymmetry of GRF in anterior-posterior and vertical directions (Van Gheluwe and Madsen, 1997). Moreover, it can be seen from Figure 6 that the symmetry of medial-lateral direction of GRF deteriorates during the push-off phase of gait (86%–88% and 90%–92%) after fatigue intervention, suggesting that the more medial load was concentrated in the metatarsal joint of non-dominant foot (Gao et al., 2020a). The deterioration of this asymmetry may be related to the risk of overuse injury to the unilateral metatarsal-toe joint, possibly due to a higher susceptibility to fatigue in the lower limb muscles of the non-dominant limb, but further research is needed to verify if GRF measures could be used to infer foot neuromuscular control.

Sensitivity and specificity were used to measure the ability of the classifier to detect gait patterns of left and right limbs and before and after fatigue, respectively (Chan et al., 2002). Earlier, Chan et al. (Chan et al., 2002) reported 91% and 94% similarity success rates for second-order POLY and linear nuclei in an SVM gender classification task based on gait video sequence data. Our results also show that the SVM model can map the underlying data structures related to asymmetrical and fatigue gait (Atanassov et al., 2021). Machine learning-based classifiers can automatically recognize particular gait patterns according to their measurement methods, which is expected to provide a basis for exploring the potential biomechanical mechanism of running-fatigued. Early

recognition of gait problems caused by symmetry changes by a machine classifier can avoid fatigue through a motor intervention program in advance, thus reducing the incidence of injuries caused by fatigue and asymmetry gait. Compared with traditional symmetry evaluation methods, the SVM algorithm based on non-linear kernel function mapping data to high-dimensional space classification has a more substantial symmetry quantization ability (Wu and Wu, 2015).

In this study, it is observed that the adjustment of optimal parameters is also crucial to the improvement of model accuracy, especially the choice of penalty parameter C and kernel function. The change of each kernel will lead to a change in the model. Therefore, the optimal model is obtained through repeated testing of a large number of experiments in this study (Begg and Kamruzzaman, 2005). The selection of kernel function has an important influence on the generalization ability of the SVM model since the kernel function reflects the internal changes of gait biomechanical characteristics by mapping the mutual non-linear relationship between gait variables into the high-dimensional feature space (Schölkopf et al., 2002). Therefore, three kernel functions (RBF, POLY, and LINEAR) were selected for gait data analysis, considering that all gait parameters may have probability distributions in higher space. It can be seen from Tables 2, 3 that the generalization performance of non-linear kernel functions such as POLY and RBF is better than that of LINEAR in the SVM model of gait feature recognition. A recent study reported by Xiang and his colleagues (Xiang et al., 2022b) showed that the SVM model based on RBF had a prediction accuracy of 93% to classify dynamic plantar pressure and foot metrics of barefoot and shod people. This finding is consistent with the results obtained in this study that the prediction accuracy of SVM model is higher than that of the LINEAR. Specifically, the POLY kernel can make the SVM model the most accurate in classifying left and right gait features, with accuracy of 85.3%, 82.4%, and 82.4% in medial-lateral, anterior-posterior and vertical directions, respectively. Similarly, RBF was the optimal kernel function of the SVM model for fatigue gait recognition in medial-lateral (54.2%) and vertical direction (62.5%). Furthermore, POLY was the optimal kernel function of the anterior-posterior direction (54.2%). Using these SVM model to achieve automatic gait feature of bilateral lower limbs and of fatigue gait classification, suggesting that early recognition runners gait asymmetry and fatigue gait. As a final summary, SVM classifier within the healthy individuals of this study provides a basis for further exploring the automatic recognition methodology of gait asymmetry and of fatigue gait.

There are also four limitations to this study. Firstly, in the process of GRF data collection, we only used one force plate to measure the left and right gait characteristics, respectively. secondly, the participants in this study are all young groups, so the model may not be applicable to older runners. Future studies should consider the biomechanical data sets of all ages for model training. In addition, although the running-induced fatigue experiment is a classic fatigue method, it is performed on a treadmill and may differ from long-distance running on the ground, such as a marathon. Moreover, the only GRF data was considered as a variable for gait recognition in this study, more variables that can sensitively reflect asymmetric gait, such as joint Angle, should be included in the further study. At last, this study selected participants' comfortable running speed for ground data collection, thus ignoring the possible effect of speed on GRF. Future studies should compare the GRF differences between comfortable and standard running speeds.

5 Conclusion

In this study, the asymmetry degree and changes of GRF in both lower limbs of 14 amateur male runners were investigated before and after a running-induced fatigue experiment. An SVM machine learning model was established to mine and recognize the characteristics of left and right gait and fatigue gait non-linear. The findings of this study suggest that GRF asymmetry existed in the medial-lateral, anterior-posterior and vertical directions of bilateral lower limbs, especially in medial-lateral direction. In addition, the asymmetry of GRF in the medial-lateral direction was increased after fatigued. Moreover, the POLY and RBF kernel contribute more to recognizing asymmetric and fatigue gait characteristics in SVM machine learning model, respectively.

Data availability statement

The raw data supporting the conclusion of this article will be made available by the authors, without undue reservation.

Ethics statement

The studies involving human participants were reviewed and approved by Ningbo University. The patients/participants provided their written informed consent to participate in this study.

Author contributions

ZG, YZ, YF, GF, and AK conceived the presented idea, developed the framework, and wrote the manuscript. JB, ML, and YG provided critical feedback and contributed to the final version. All authors were involved in the final direction of the paper and contributed to the final version of the manuscript and have read and agreed to the published version of the manuscript.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Effects of mental fatigue on biomechanical characteristics of lower extremities in patients with functional ankle instability during unanticipated side-step cutting

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Background: Functional ankle instability (FAI) is the primary classification of ankle injuries. Competitive activities have complicated movements that can result in ankle re-injury among patients with FAI. Unanticipated movement state (MS) and mental fatigue (MF) could also happen in these activities, which may further increase their joint injury risk.

Objective: This study aimed to clarify the biomechanical characteristics difference of the lower extremity (LE) between the injured side and the uninjured side among patients with FAI when they perform unanticipated side-step cutting after MF.

Methods: Fifteen males with unilateral FAI participated in this study (age: 20.7 ± 1.3 years, height: 173.6 ± 4.4 cm, weight: 70.1 ± 5.0 kg). They used the injured side and the uninjured side of LE to complete anticipated and unanticipated side-step cutting before and after MF. The kinematic and kinetics data were evaluated using three-way ANOVA with repeated measures.

Results: During patients with FAI performed anticipated side-step cutting, the ankle stiffness of both sides showed no significant change after MF; During they performed unanticipated side-step cutting, their injured side presented significantly lower ankle stiffness after MF, while the uninjured side did not have such change. In addition, after MF, the injured side exhibited increased ankle inversion, knee valgus and LR, but the uninjured side did without these changes.

Conclusion: Influenced by MF, when patients with FAI use their injured side of LE to perform side-step cutting, this side LE has a higher risk of musculoskeletal injuries such as lateral ankle sprains and anterior cruciate ligament injury. The ankle stiffness of the injured side will be further reduced when patients with FAI perform unanticipated side-step cutting, which increases ankle instability and the risk of re-injury.

KEYWORDS

mental fatigue, functional ankle instability, side-step cutting, movement state, ankle stiffness

1 Introduction

Lateral ankle sprain is the most common type of sports injury in the lower extremity (LE), with an incidence rate of approximately 7.3%; it is mostly caused by abrupt excessive ankle inversion (Roos et al., 2017). If the injured ankle is not properly treated after the first sprain, it will continue to be negatively affected by chronic pain and local edema. Patients might be unable to control the injured ankle sufficiently, increasing its re-injury risk and eventually leading to functional ankle instability (FAI) (Cao et al., 2019).

FAI is the primary classification of ankle instability in clinical, and it can be attributed to the loss of proprioception and neuromuscular deficits in the injured ankle (Steib et al., 2013). FAI is typically accompanied by abnormal joint sensation and loss of normal motion control of the ankle (Freeman et al., 1965; Marinho et al., 2017). The abnormal joint sensation usually occurs in patients with FAI (Freeman et al., 1965; Arnold et al., 2009; Lysdal et al., 2022), which refers to frequently unconscious episodes of hyper-inversion and subsequent "giving way" of the ankle likely to occur (Tropp, 2002; Takeda et al., 2021). Such abnormal ankle motion may aggravate the risk of ankle re-injury in patients with FAI during sports. Side-step cutting is one of the complex movements commonly used in competitive activities, and its main objective is to dodge the defensive player through sudden direction changes while running (Besier et al., 2001). This movement usually generates an impact on the ankle joint that is three times more than one's body weight, and the generated instantaneous pressure is largely borne by the LE. Recent studies have shown that patients with ankle instability exhibit significant biomechanical differences in side-step cutting performance compared with healthy individuals, especially the significantly greater ankle internal rotation that may result in ankle re-injury (Simpson et al., 2020a; Simpson et al., 2020b).

When humans perform movements, they need to fully account for the external environment, and follow-up actions will be preplanned under the current movement state (MS) (Dey and Schilling, 2022). MS in competitive sports is constantly changing. Participants do not know the direction or route of action in advance and need to make immediate adjustments based on the instant feedback from the sensory system (Besier et al., 2001). Sudden unanticipated disturbances can evoke psychological and physiological responses to such stimulus, namely, the startle reflex. The main function of the startle reflex is to help the body avoid external stimulation and increase the sympathetic nervous system activity to prepare for subsequent actions (Yeomans et al., 2002). But at the same time, because the startle reflex is an autonomous defensive reflex and is not controlled by the will, its appearance is likely to lead to the deformation of the standard action. In unanticipated MS, the human body cannot quickly and accurately make corresponding postural adjustments like anticipated MS, resulting in involuntary changes in neuromuscular activities. Some scholars pointed out that unanticipated side-step cutting increases the knee valgus angle and the ligament injury risk (Brown et al., 2014). Patients with FAI have a diminished capacity to use their injured side of LE to maintain balance. They cannot adapt to the external environment and adjust their posture in time (Kazemi et al., 2017). They might be more likely to suffer from injury when performing unanticipated side-step cutting. Analyzing the biomechanical characteristics of these patients who perform unanticipated side-step cutting can provide a theoretical basis and references for preventing sports injury.

Patients with ankle instability experience difficulty maintaining postural balance after prolonged and intensive exercise due to proprioceptive deficits and weak ankle motion control (Gribble et al., 2007), making them more at risk of excessive ankle torsion. Both mental fatigue (MF) and muscle fatigue caused by long-term exercise could reduce sports performance quality. But different from muscle fatigue, MF is a psychobiological state caused by prolonged periods of demanding cognitive activity (Van Cutsem et al., 2017; Le Mansec et al., 2018; Meeusen et al., 2021), emphasizing the difficulty of CNS signal integration and the increased cognitive burden (Li et al., 2015). Previous studies have revealed that individuals are unable to concentrate and maintain efficiency in working after MF (Ream and Richardson, 1996) and also have difficulty changing coping strategies in the face of the external environment or in performing movement tasks (Lorist et al., 2005; Pageaux and Lepers, 2018). After moderate-to low-intensity aerobic exercise, jeopardized behavioural and cognitive control induced by MF disables the general muscle contraction even though muscles are without fatigue (Olson et al., 2016). Although both MF and MS have been proven to be potential factors affecting sports safety, most studies only analyzed the movement performance of patients with FAI when they complete the anticipated side-step cutting (Dayakidis and Boudolos, 2006; Suda and Sacco, 2011). A few studies have analyzed the biomechanical performance of these patients while they perform the unanticipated step-cutting task (Kim et al., 2021a; Kim et al., 2021b), but scholars have not further considered the possible effects of MF in this state.

The current study is aimed to clarify the biomechanical characteristics of the LEs of patients with FAI during anticipated or unanticipated side-step cutting before and after MF. We hypothesized the following: 1. The patients with FAI show different kinematic and kinetic characteristics between the injured and the uninjured sides during side-step cutting; 2. The appearance of MF or unanticipated movement affects the kinematic and kinetic characteristics of the injured side during side-step cutting; 3. Influenced by MF, the injured side could show obvious biomechanical characteristics changes during these patients perform unanticipated side-step cutting in comparison with the uninjured side.

2 Materials and methods

2.1 Participants

Fifteen males with unilateral FAI (age: 20.7 ± 1.3 years, height: 173.6 ± 4.4 cm, weight: 70.1 ± 5.0 kg) who frequently participated in sports (i.e., at least three times a week) and had good side-step cutting skills, were recruited for this study. The dominant LE of all the participants was right determined by kicking a ball; it was also the injured side. The inclusion criteria for patients with FAI were as follows. (1) The unilateral ankle had suffered at least one sprain in the past year, and the patient experienced a feeling of weakness or instability. (2) The score in the Cumberland Ankle Instability Tool was below 24 (Donahue et al., 2011); (3) The patient had no critical

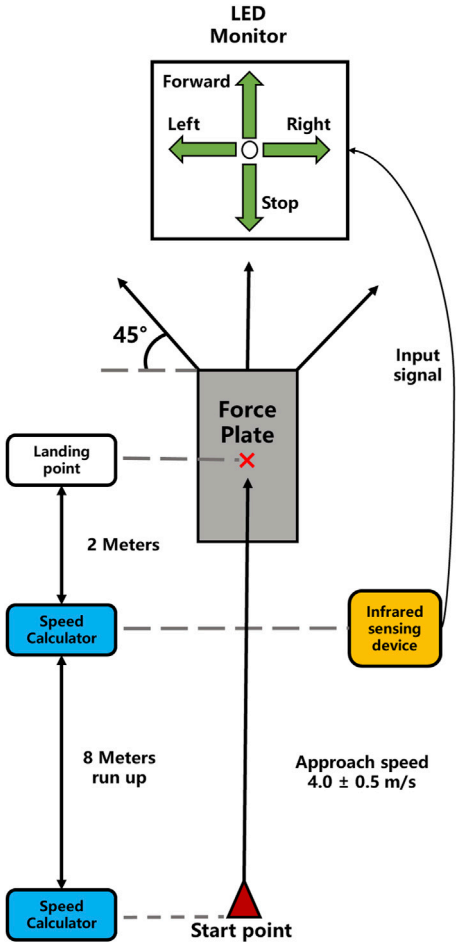

FIGURE 1
Layout of experimental site.

LE injury history, including fractures or serious orthopedic injury (Wu et al., 2022), except for ankle sprain; (4) The anterior drawer test and the talar tilt test were negative (Kaminski et al., 1999); (5) Only one LE side suffered from FAI. The exclusion criteria were as follows: (1) Bilateral ankle sprains (Wang et al., 2022); (2) Acute pathological symptoms of LE; (3) History of previous surgeries in the LE (Kweon et al., 2022); (4) History of equilibrium and balance control disorder (Wu et al., 2022); (5) Had congenital feet, ankles, knees, pelvis and spine deformities. This study was approved by Ethics Committee of Soochow University, and all the participants had written consent forms before the experiment.

2.2 Side-step cutting

The design of the side-step cutting task followed that of previous research (Kim et al., 2014; Liew et al., 2021). In summary, a minimum of 8 m run-up distance was provided to participants to reach the desired speed (4.0 ± 0.5 m/s). Participants stepped their foot on the fixed position of the force platform (90 cm × 60 cm × 10 cm) in the mode of rear foot landing. Then they quickly ran away from the force platform towards the cutting direction. When participants completed the left-direction side-step cutting, they must complete the cutting task by the right side of LE. On the contrary, they needed to use the left side of LE during the cutting direction is right. All participants were required to run at least three steps before decelerating and stopping.

Side-step cuttings were performed under anticipated and unanticipated states in sequence. A light-emitting diode (LED) monitor with four green bulbs in the arrow shape was set in behind the force platform to give the side-step cutting direction order. The left (right) arrow represented participants should perform maximum-effort side-step cutting to the left (right) 45° direction. The upper arrow representative continued to run forward, while the lower arrow representative emergently stopped. For the anticipated state, the bulb with the left (or right) arrow has lit before participants run up. For the unanticipated state, the LED monitor was connected to the infrared sensing device. Only when the participants passed the infrared sensing device was one of the four arrow bulbs applied through a computer program randomly lit. The layout of experimental site is shown in Figure 1.

2.3 Induction and assessment of MF

The Stroop task has been proven effective in inducing MF (Mangin et al., 2022). In the current study, a 45-min Stroop task included four words in Chinese (red, yellow, blue, and green) that were displayed randomly in one of the four colors. The task was presented in an event-related design. It comprised 675 trials that included 225 congruent trials (the meaning of the color matched the color) and 450 incongruent trials (the meaning of the color did not match the color). Each trial began with a 500 ms fixation cross, followed by the stimulus presented for 2 s and then the blank for 1.5 s. The subjects were required to press one of the four keys to indicate the color of the ink while disregarding the meaning of the color word.

As a complex probe of attention that combines both visual selectivity and motor response, letter cancellation task (LCT) can reflect the participants' changes in attention, attention span, and vigilance (Casagrande et al., 1997; Casagrande et al., 1999; Pradhan et al., 2018), and can be used in neurological status assessment (Geldmacher, 1998). In this study, the degree of participants' MF before and after receiving the 45-min Stroop task were assessed by LCT. Participants were required to search for and mark target letters as quickly and accurately as possible (Casagrande et al., 1997). Target letters were randomly placed within a 20 × 53 matrix of capital letters (font: Time New Roman, size: 12) printed on an A4-size paper. Participants were ordered to complete LCT as soon as possible. Different matrices of capital letters were used to analyze the changes in the level of MF in each participant before and after they received the Stroop task. Whether the changes in the performance of LCT in participants were significantly worsened were used to determine their appearance of MF.

2.4 Experimental procedure

All the participants were asked to refrain from strenuous exercises 24 h before the formal test. Before data collection, they performed a warm-up exercise that comprised of a 5-min jogging on a treadmill at their preferred speed (Zebris FDM-T, Germany).

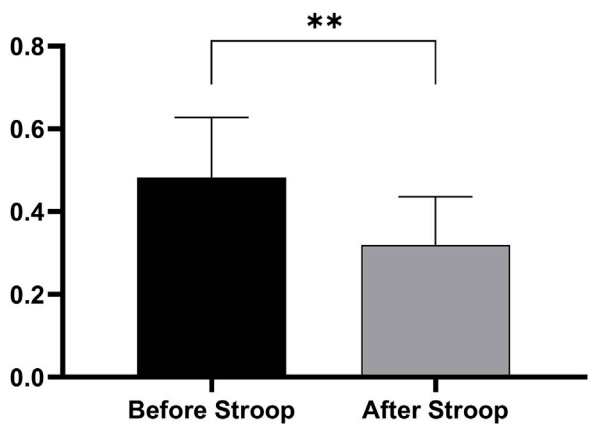

FIGURE 2
Changes in work efficiency of LCT in participants before and after Stroop task.

All participants first completed three successful trials of side-step cutting movement under both anticipated and unanticipated conditions, performed the Stroop task to induce MF, and then repeated three successful trials of the same side-step cutting movement under the two testing conditions. Data were collected and recorded during each trial. A successful trial was defined as a run-up speed within 4.0 ± 0.5 m/s. Side-step cutting should realize the correct footfalls and land on the point. The cutting angle must be within 45° ± 3°. The mean value of three trials was used for comparative analysis. Finally, the successful 45° side-step cutting of the uninjured side and injured side under different MSs before and after MF was analyzed and compared. Running forward and emergent stops were used as interference items under unanticipated MS, and their trial data were not considered in the subsequent analysis and comparison.

2.5 Data processing

Kinematic data were captured using a motion analysis system that comprised eight infrared cameras (Vicon Motion Analysis, United Kingdom) by tracking 16 infrared reflective balls (reflective markers) with a diameter of 14 mm at 100 Hz. The infrared reflective balls were stick to participants' corresponding parts following the scheme suggested by the Plug-in Gait model. Kinetic data were capture using a 3D force plate (KISTLER, Switzerland) at 1,000 Hz, which was synchronized with motion analysis system. Kinematic and kinetic data were firstly processed by Vicon Nexus 2.1.2. Both kinematic and kinetic data were then imported to Visual3D (Version 6, C-Motion, Inc, United States) for further processing. The following data were analyzed: peak joint angles on the sagittal and frontal planes (Kim et al., 2014); ground-reaction force (GRF), including peak vertical GRF (vGRF), peak medial GRF (mGRF), peak horizontal GRF (hGRF); time-to-peak ground reaction force (T_GRF), including time-to-peak vertical GRF (T_vGRF), time-to-peak medial GRF (T_mGRF), time-to-peak horizontal GRF (T_hGRF), and stance duration; ankle stiffness and loading rate (LR). GRF data were standardized by each participant's body weight (BW).

Working efficiency can comprehensively evaluate the performance of participants when completing LCT (Yang, 1989). We used working efficiency to analyze the fatigue degree of the central nervous system (CNS) to determine whether MF was successfully induced. Working efficiency was calculated using Equation 1.

$$A = \frac{(c - w)}{(c + o)}$$
$$E = 100*A/T \quad (1)$$

A denotes the accuracy of cancellation, c represents the number of cancelled symbols, w represents the number of wrongly cancelled symbols, o represents the number of missed cancelled symbols, T is the time taken to cancel symbols, and E indicates working efficiency.

LR can be raised due to load accumulation caused by abnormal movement patterns, which are closely associated with injury. LR was calculated using Equation 2.

$$LR = vGRF/T_vGRF \quad (2)$$

Ankle joint stiffness is the ratio of the change of ankle joint moment and ankle joint angular displacement from the moment of touchdown to the moment of maximum ankle dorsiflexion (Kim et al., 2019). In the current study, we used the ankle stiffness of different LEs to quantify the interaction between MF and MS. Ankle stiffness was normalized to each participant's BW. Ankle stiffness was calculated using Equation 3.

$$K_{ankle} = \Delta M_{ankle}/\Delta\theta_{ankle} \quad (3)$$

ΔM is defined as the change in ankle moment, and Δθ represents the angular displacement in ankle dorsiflexion (Hamill et al., 2014).

2.6 Statistical analysis

Data were expressed as mean and standard deviation (SD). SPSS 26.0 software (SPSS Inc, Chicago, IL, United States) was used for statistical analysis. Normality tests were conducted *via* the Kolmogorov–Smirnov test. Matched sample *t*-test was used to assess the working efficiency of the participants before and after the Stroop task. A three-way (2 LE × 2 MF × 2 MS) analysis of variance with repeated measures was performed for kinematic and kinetic variables. Interaction effects will be investigated prior. Main effects were considered only if non-significant interaction effects were found. In case a significant interaction was detected, simple effects analysis was performed (Keppel et al., 1992). Statistical significance was set at 0.05 for all variables.

3 Results

Figure 2 displays the results of the change in the working efficiency of LCT before and after the Stroop task. All the participants demonstrated a remarkable decrease in work efficiency after the Stroop task (t = 3.097, p = 0.008), indicating that MF was successfully induced.

TABLE 1 Maximum Angles of the hip, knee, and ankle joints during side-step cutting.

Variables (+/–)	Injured side		Uninjured side		Main effect			Interaction effect			
	Anticipated movement	Unanticipated movement	Anticipated movement	Unanticipated movement	LE	MF	MS	LE × MF	LE × MS	MF × MS	LE × MF × MS
Ankle dorsiflexion/plantarflexion (degree)											
Before MF	18.8 (5.6)	19.2 (5.7)	20.2 (5.9)	23.5 (6.1)	**0.015***	**0.002****	**0.013***	0.971	0.574	0.424	0.318
After MF	20.1 (4.8)	23.7 (4.9)	23.3 (4.3)	26.2 (4.0)							
Ankle eversion/inversion (degree)											
Before MF	−10.5 (5.8)	−10.3 (3.1)	−10.2 (6.2)	−10.6 (4.6)	0.118	0.202	0.296	**0.040***	0.608	0.300	0.358
After MF	−11.6 (5.9)	−15.1 (7.4)	−9.3 (5.0)	−10.0 (4.2)							
Knee flexion/extension (degree)											
Before MF	45.7 (5.1)	51.9 (6.4)	47.6 (8.0)	51.5 (6.4)	0.681	0.971	**0.001****	0.864	0.984	0.975	0.480
After MF	47.1 (7.5)	51.1 (5.3)	46.2 (11.2)	52.5 (8.8)							
Knee valgus/varus (degree)											
Before MF	3.7 (5.0)	4.5 (3.1)	3.6 (6.2)	4.3 (4.1)	**0.043***	**0.032***	0.057	**0.043***	0.333	0.434	0.417
After MF	6.1 (4.6)	9.7 (4.5)	3.7 (5.0)	4.4 (5.2)							
Hip flexion/extension (degree)											
Before MF	42.9 (13.7)	49.2 (14.4)	48.6 (9.6)	50.3 (9.3)	0.210	0.973	0.098	0.971	0.811	0.684	0.341
After MF	45.5 (11.2)	46.6 (15.3)	47.4 (9.9)	51.2 (4.9)							
Hip abduction/adduction (degree)											
Before MF	4.3 (3.2)	6.6 (7.6)	4.5 (7.2)	8.1 (4.8)	0.803	0.351	**0.025***	0.648	0.146	0.877	0.538
After MF	4.0 (6.0)	5.7 (4.6)	6.8 (6.1)	7.0 (5.7)							

"*" means significance ($p< 0.05$), "**" means very significance ($p< 0.01$). Abbreviations: LE: lower extremity, MF: mental fatigue, MS: movement state.
Bold value is the appearance of statistical significance.

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TABLE 2 Peak ground reaction force, time to peak ground reaction force, and stance duration during side-step cutting.

Variables	Injured side		Uninjured side		Main effect			Interaction effect			
	Anticipated movement	Unanticipated movement	Anticipated movement	Unanticipated movement	LE	MF	MS	LE × MF	LE × MS	MF × MS	LE × MF × MS
vGRF (BW)											
Before MF	3.20 (0.75)	2.76 (0.79)	3.38 (0.65)	2.63 (0.71)	0.463	0.593	**0.003****	0.372	0.436	0.229	0.741
After MF	3.27 (0.51)	3.09 (0.91)	3.09 (0.94)	2.81 (0.75)							
mGRF (BW)											
Before MF	0.96 (0.29)	0.71 (0.17)	0.83 (0.21)	0.73 (0.18)	0.892	0.330	**0.000****	0.136	0.060	0.476	0.589
After MF	0.88 (0.14)	0.62 (0.16)	0.90 (0.10)	0.70 (0.22)							
hGRF (BW)											
Before MF	1.06 (0.26)	1.18 (0.22)	1.21 (0.21)	1.23 (0.20)	**0.002****	0.836	0.509	0.593	0.474	0.418	0.784
After MF	1.08 (0.25)	1.10 (0.29)	1.25 (0.23)	1.21 (0.26)							
T_vGRF (s)											
Before MF	0.049 (0.011)	0.043 (0.015)	0.055 (0.011)	0.042 (0.011)	0.375	0.321	**0.003***	0.976	0.148	0.196	0.906
After MF	0.044 (0.008)	0.043 (0.014)	0.049 (0.017)	0.043 (0.012)							
T_mGRF (s)											
Before MF	0.226 (0.053)	0.212 (0.077)	0.254 (0.054)	0.208 (0.057)	0.391	0.092	0.586	0.977	0.162	0.079	0.980
After MF	0.223 (0.038)	0.256 (0.083)	0.249 (0.089)	0.252 (0.070)							
T_hGRF (s)											
Before MF	0.095 (0.023)	0.101 (0.037)	0.105 (0.021)	0.097 (0.027)	0.812	0.089	0.563	0.775	0.397	0.751	0.708
After MF	0.110 (0.020)	0.107 (0.037)	0.112 (0.041)	0.105 (0.029)							
Stance duration (s)											
Before MF	0.379 (0.087)	0.387 (0.139)	0.392 (0.083)	0.378 (0.102)	0.907	0.096	0.502	0.987	0.610	0.495	0.941
After MF	0.399 (0.070)	0.434 (0.140)	0.409 (0.144)	0.429 (0.119)							

"*" means significance ($p < 0.05$), "**" means very significance ($p < 0.01$). Abbreviations: LE: lower extremity, MF: mental fatigue, MS: movement state.
Bold value is the appearance of statistical significance.

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TABLE 3 Ankle stiffness and lower extremities loading rate during side-step cutting.

Variables	Injured side		Uninjured side		Main effect			Interaction effect		
	Anticipated movement	Unanticipated movement	Anticipated movement	Unanticipated movement	LE	MF	MS	LE × MF	LE × MS	MF × MS	LE × MF × MS
K_{ankle} (N·m/kg/°)											
Before MF	0.051 (0.012)	0.049 (0.009)	0.054 (0.009)	0.050 (0.007)	**0.011***	**0.000****	**0.003****	0.139	**0.040***	0.173	**0.046***
After MF	0.049 (0.011)	0.032 (0.010)	0.048 (0.011)	0.047 (0.011)							
LR (BW/s)											
Before MF	65.08 (4.33)	65.96 (5.37)	61.99 (6.00)	63.32 (5.35)	**0.001****	**0.000****	0.834	**0.001****	0.163	0.314	0.372
After MF	74.74 (5.87)	72.05 (6.75)	64.33 (6.13)	65.44 (4.69)							

"*" means significance ($p < 0.05$), "**" means very significance ($p < 0.01$). Abbreviations: LE: lower extremity, MF: mental fatigue, MS: movement state.
Bold value is the appearance of statistical significance.

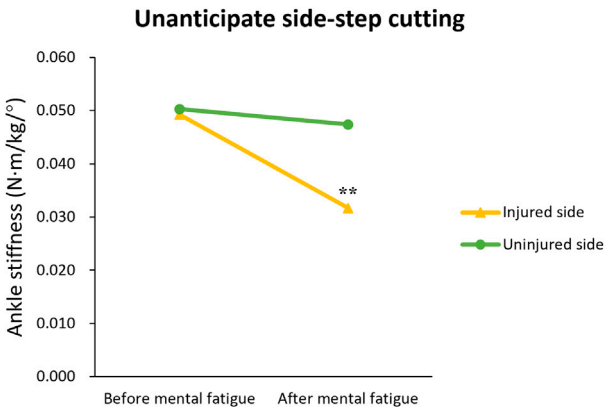

FIGURE 3
Change trend in ankle stiffness among patients with functional ankle instability using the injured and the uninjured sides lower extremities perform unanticipated side-step cutting before and after mental fatigue. "**" means a very significant statistical difference compared with before mental fatigue ($p < 0.01$).

Table 1 presents the results in comparison of the peak joint angles between the injured side and the uninjured side. LE (F = 6.704, p = 0.015, η^2 = 0.193), MF (F = 11.512, p = 0.002, η^2 = 0.291), and MS (F = 7.100, p = 0.013, η^2 = 0.202) showed significant main effects on ankle dorsiflexion. LE × MF exhibited a significant interaction effect on ankle inversion (F = 4.624, p = 0.040, η^2 = 0.142). The simple effect analysis showed that the injured side had significantly increased ankle inversion after MF (p = 0.021); However, the uninjured side did not have a similar change (p = 0.566). MS showed a significant main effect on knee flexion (F = 13.486, p = 0.001, η^2 = 0.325). LE (F = 4.475, p = 0.043, η^2 = 0.138), and MF (F = 5.094, p = 0.032, η^2 = 0.154) showed significant main effects on knee valgus. LE × MF showed a significant interaction effect on knee valgus (F = 4.516, p = 0.043, η^2 = 0.139). The simple effect analysis showed that the injured side had significantly increased knee valgus after MF (p = 0.003), while the uninjured side did not exhibit such a change (p = 0.926). MS exhibited a significant main effect on hip abduction (F = 5.607, p = 0.025, η^2 = 0.167).

Table 2 presents the results in comparison of GRF, T_GRF and stance duration between the injured side and the uninjured side. MS showed significant main effects on vGRF (F = 10.313, p = 0.003, η^2 = 0.269), mGRF (F = 55.882, p < 0.001, η^2 = 0.666), and T_vGRF (F = 10.485, p = 0.003, η^2 = 0.272). LE presented a significant main effect on hGRF (F = 11.198, p = 0.002, η^2 = 0.286). No interaction effects were observed in this table.

Table 3 provides the results in comparison of ankle stiffness and loading rate between the injured and the uninjured side. LE (F = 7.523, p = 0.011, η^2 = 0.212), MF (F = 17.738, p < 0.001, η^2 = 0.388) and MS (F = 10.177, p = 0.003, η^2 = 0.267) have significant main effects on ankle stiffness; LE × MS (F = 4.659, p = 0.040, η^2 = 0.143), and LE × MF × MS (F = 4.373, p = 0.046, η^2 = 0.135) have significant interaction effects on ankle stiffness. The simple effect analysis of LE × MF × MS showed that during anticipated cutting, both sides showed no significant change in ankle stiffness after MF (p = 0.752; p = 0.399); During unanticipated cutting, the injured side presented

significantly decreased ankle stiffness after MF ($p < 0.001$), whereas the uninjured side did not exhibit such change ($p = 0.963$) (Figure 3).

LE ($F = 14.588$, $p = 0.001$, $\eta^2 = 0.343$) and MF ($F = 43.942$, $p < 0.001$, $\eta^2 = 0.611$) showed significant main effects on LR. LE × MF exhibited a significant interaction effect on LR ($F = 13.769$, $p = 0.001$, $\eta^2 = 0.330$). The simple effect analysis showed that the injured side had higher LR after MF ($p < 0.001$), but the uninjured side did not show changes ($p = 0.069$).

4 Discussion

In this study, we found the biomechanical characteristics difference of the lower extremity (LE) between the injured side and the uninjured side among patients with FAI when they performed side-step cutting. This finding approved our hypothesis 1; Mainly MF but not MS would affect these patients' biomechanical characteristics of LE. This finding partly approved our hypothesis 2; Consistent with our hypothesis 3, we observed that during unanticipated side-step cutting, the ankle stiffness of the injured side decreased significantly after MF. Since we observed interaction effects, we mainly discussed these findings next.

Consistent with the previous study, MF induced by the Stroop test negatively affected the kinematic performance among elite sporters (Veness et al., 2017); our results showed that MF induced by the Stroop test influenced the normal motion control of patients with FAI in their ankle and knee in the front plane of the injured side. The side-step cutting movement has high requirements for the non-sagittal movement of the LE when landing (Zhou et al., 2021). The function of muscles in controlling the ankle in patients with FAI is impaired (Delahunt et al., 2006), and Koshino et al. (2016) found that the ankle inversion angle of patients with ankle instability was significantly higher than that of healthy individuals when completing side-step cutting. Our results are consistent with them and showed that the injured side had increased ankle inversion compared to the uninjured side after MF. If the ankle inversion of patients with ankle instability is reduced while performing 45° side-step cutting, representing they take prudent and conservative sports strategies to avoid ankle injury (Son et al., 2017; Fuerst et al., 2018). For patients with FAI, they must exert more energy to make their CNS consciously control ankle stability. Reduced CNS control caused by MF inhibits such protection and finally increases ankle inversion. Similarly, Pageaux et al. (2014) found that 30 min of mental exertion involving response inhibition reduces subsequent self-paced endurance performance, negatively affecting the normal motion control of LE. Increased ankle inversion indicates that the lateral ankle musculature of patients with FAI is unable to control frontal-plane motion eccentrically when the lateral ankle is loaded during ground contact, which leads to the ankle complex giving way to excessive inversion and exposing the injured side to an increased risk of recurrent lateral ankle sprain during side-step cutting (Simpson et al., 2020b).

Our results also showed that the injured side had increased knee valgus compared with the uninjured side after MF. This result is in line with the research view of David et al. (2017), that is, the increased knee valgus will better meet the needs of people who suffer from ankle injuries. In the current study, all patients with FAI were ordered to adopt the rear foot landing mode. vGRF of the rear foot landing is 3.4 times that of the front foot landing mode (Kovács et al., 1999), and the knee joint turn outward can better absorb vGRF (Mizuno et al., 2009) to reduce the ankle needs to bear when performing movements. The increased knee valgus of the injured side proved that the appearance of MF affected and adjusted the GRF distribution of LE, reducing the ankle burden. However, Son et al. (2017) have found that although elastic ankle protection can reduce the energy absorption of the ankle, which protects the damaged ankle joint, the negative impact still has on the knee joint. Most patients with ankle instability reported more symptomatology in the knee and worse knee joint health than healthy individuals (Kosik et al., 2020). Increased knee valgus can induce anterior cruciate ligament injury during side-step cutting (McLean et al., 2005; David et al., 2017). Hence, influenced by MF, patients with FAI rely more on changing knee motion to cushion the GRF of side-step cutting, but this further aggravates the risk of knee joint injury.

In this study, the injured side showed higher LR after MF, consistent with Tajdini et al. (2022) that patients with ankle instability exhibited a greater inter-limb asymmetry of LR, and the LR of their injured side was higher than healthy individuals during walking. As we found in kinematics, the FAI population after MF cannot effectively control the ankle inversion angle. Greater ankle movement control can decrease LR (Decker et al., 2002), reducing the impact of stress on soft tissues during landing. The impaired ability in injured ankle motion control of patients with FAI might be the main reason for the high LR. Besides, some scholars (Le Mansec et al., 2018) pointed out that MF directly affects the depth and speed of visual processing before attention. These changes negatively affect the precision and integrity of automatic visual processing, influencing the subsequent concentrated attention stage processing and ultimately leading to error task execution (Rozand et al., 2015; Smith et al., 2016). In side-step cutting, the human body must brake to reduce the forward speed when touching the ground first and then transfer part of the forward speed to the side speed by pushing the ground in the opposite direction of the side cut, accelerating the push to achieve the purpose of a side cut. All the actions mentioned above should be completed quickly. Due to the damaged muscle spindle and around receptors of the injured ankle, nerve signal transmission speed from the joint to the CNS was directly impeded. The appearance of MF will further negatively influence the signal process of CNS, and sensory information will not be processed in time. These patients adopt an uncomplicated buffering strategy to complete side-step cutting, which requires less time, and posture preparation process of the injured side, but soft tissues or muscles of the injured side of LE are not fully mobilized to cushion the load during side-step cutting, making raised LR, which will cause stress fracture and plantar fasciitis (Venesky et al., 2006).

When the posture of the human body changes, the adjustment of LE stiffness is the first thing to start (Liu et al., 2006). If joint stiffness cannot effectively adjust the impact and influence generated by MS, it will inevitably reduce the quality of movement performance, such as the deformation of LE movements during landing (Flanagan and Harrison, 2007; Balasundaram and Rajan, 2018), leading to musculoskeletal injury. During the anticipated MS, humans can prepare enough to adjust the physical characteristics of soft tissues by actively activating muscles, the neuromuscular system will adjust muscle tuning according to the MS (Nigg and Wakeling, 2001), and the stiffness and motion of joints will also be adjusted accordingly to activate the joint soft tissue and reduce the risk of injury (Nigg and Liu, 1999). Increased ankle stiffness can be used to stabilize the body and prevent excessive joint motion (Brughelli and Cronin, 2008; Li

et al., 2021). In this study, MF had no significant influence on ankle stiffness of the injured side when patients with FAI performed anticipated side-step cutting. Consistent with our results, Simpson et al. (2020a) found that less lateral center of pressure progression and increased tibialis anterior activation were observed in patients with ankle instability, reflecting a protective movement strategy during anticipated side-cutting to avoid recurrent injury. Therefore, when patients with FAI use their injured side to perform the anticipated side-step cutting, the appearance of MF could not result in an obviously negative influence on the ankle joint.

However, the amount of time for participants to make appropriate postural adjustments before performing the unanticipated cutting task is smaller than the anticipated state (Besier et al., 2001), and their movement plan must be immediately adjusted in the CNS based on current MS. In this process, the thinking decision is dominant in their brain, but their movement control is inevitably weakened. The appearance of MF will further delay the thinking decision process, leading to abnormal action control (Pageaux et al., 2014). We found that, influenced by MF, the injured side had significantly decreased ankle stiffness when these patients performed unanticipated side-step cutting. This finding is consistent with Kim et al. (2019) that the ankle stiffness of patients with ankle instability is lower than that of healthy individuals. Brown et al. (2022) revealed that the decreased ankle stiffness of the injured side indicates its diminished ability to respond to movement loading. A compliant joint contributes more to the attenuation of the joint load than a stiffer joint (Hamil et al., 2014), and such altered neuromechanics in patients with ankle instability means redistributing energy absorption from the distal (ankle) to the proximal (knee and hip) joints (Kim et al., 2019). The ankle stiffness change might be a protection strategy used by patients with FAI (Sarvestan et al., 2021) to modulate impact forces during this task. But due to side-step cutting being a complex and intense exercise, the decreased ankle stiffness also means their ankle stability will be weaker in this task. A previous study revealed that patients with ankle instability demonstrated alterations in landing/cutting movement strategies. These patients have a higher susceptibility to foot placement for lateral ankle sprains (Kim et al., 2019). Consequently, influenced by MF, when patients with FAI use their injured side to perform unanticipated side-step cutting, their injured ankle easily suffers from re-injury.

5 Clinical implications

This study demonstrated that compared with MS, MF is more likely to lead to patients with FAI having biomechanical characteristics changes in the injured side of LE. Proper physical exercise is very important to improve the lower limb joint control of this group of people. Balance training can improve the neuromuscular control ability of people with FAI when completing dynamic tasks, thus improving test performance (Wu et al., 2022). The findings of kinematics suggest that after MF, patients with FAI have control obstacles in sagittal control of ankle and knee joints. Progressive hop-to-stability balance (PHSB) training is a balance training method that emphasizes dynamic stability in predictable or unpredictable jump direction changes, take-off and landing plane and dynamic touch tasks. Previous studies have applied the PHSB training method proposed by McKeon et al. (2008) to train patients with ankle instability and have suggested that PHSB training is of great help in improving the posture control of people with ankle instability (McKeon et al., 2009; Anguish and Sandrey, 2018). Ardakani et al. (2019) further found that after patients with ankle instability received a 6-week hop-stabilization training program, they had better control of frontal-plane joint angles at the ankle and knee, which reduced the joint injury risk. This finding suggests that PHSB might greatly improve the kinematic performance of patients with FAI during side-step cutting and minimize the influence of MF.

Besides, influenced by MF, when patients with FAI use their injured side to perform side-step cutting, this side has higher LR than their uninjured side. Since this population relies heavily on the feedforward function to adjust posture (Delahunt et al., 2006), treatment personnel can increase their feedforward function training to maintain the posture control ability of these patients and prevent potential injuries induced by abnormal kinetic changes after MF. Some scholars have reported that gait retraining programs that utilize biofeedback can reduce high LR caused by abnormal movement patterns (Phan et al., 2017), which may be an effective training method for these patients to reduce the higher LR of the injured side. In addition, when performing unanticipated side-step cutting, the changes in ankle stiffness of the injured side ankle after MF may induce injury risk to the injured ankle. Treatment personnel should delay the emergence of MF in patients with FAI to avoid the possibility of fatigue injury to these patients' ankles. Some scholars (Chen and Zhang, 2021) have proposed that moderate but not excessive exercise can improve the level of brain metabolism and balance the secretion of neurotransmitters, improving cognition. Designing targeted exercise treatment plans for this population is worth exploring in the future.

6 Limitation

This study still has some limitations. Although this study tries to restore the performance of patients with FAI when they complete the side-step cutting in a natural sports environment and analyze their biomechanical characteristics, the number of samples is still small, which may lead to a decline in the effectiveness of some results. Besides, ankle muscles play a vital role in postural stability. Since electromyography was not used in this study, the activation of those muscles is still not apparent during the experimental process, and relevant indexes of electromyography can be considered in future research.

7 Conclusion

In conclusion, MF and MS affect the LE biomechanical characteristics of patients with FAI during side-step cutting. Influenced by MF, patients with FAI will adopt protective strategies by increasing knee valgus to buffer GRF when using their injured side of LE to complete side-step cutting. However, an increased load and a higher risk of musculoskeletal injuries such as lateral ankle sprains and anterior cruciate ligament injuries remain on this side LE. In particular, when their brain is fatigued, and patients with FAI perform unanticipated side-step

cutting using the injured side of LE, ankle stiffness considerably decreases, which enhances the possibility of bone and soft tissue damage. To improve the safety of the joints of patients with FAI during competitive sports, we suggest these patients should avoid performing unanticipated side cutting after MF. Athletic trainers and treatment personnel need to take appropriate treatment to improve these patients' cognition to delay the occurrence of MF.

Data availability statement

The raw data supporting the conclusion of this article will be made available by the authors, without undue reservation.

Ethics statement

This study was approved by the Soochow University institutional review board, and all the participants had written consent forms before the experiment.

Author contributions

LK, PW, and XZ conceived the presented idea, developed the framework, LM and LK wrote the manuscript. QZ and JS provided critical feedback and contributed to the final version. All authors were involved in the final direction of the paper and contributed.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Relationship between motor performance and cortical activity of older neurological disorder patients with dyskinesia using fNIRS: A systematic review

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Background: Neurological disorders with dyskinesia would seriously affect older people's daily activities, which is not only associated with the degeneration or injury of the musculoskeletal or the nervous system but also associated with complex linkage between them. This study aims to review the relationship between motor performance and cortical activity of typical older neurological disorder patients with dyskinesia during walking and balance tasks.

Methods: Scopus, PubMed, and Web of Science databases were searched. Articles that described gait or balance performance and cortical activity of older Parkinson's disease (PD), multiple sclerosis, and stroke patients using functional near-infrared spectroscopy were screened by the reviewers. A total of 23 full-text articles were included for review, following an initial yield of 377 studies.

Results: Participants were mostly PD patients, the prefrontal cortex was the favorite region of interest, and walking was the most popular test motor task, interventional studies were four. Seven studies used statistical methods to interpret the relationship between motor performance and cortical activation. The motor performance and cortical activation were simultaneously affected under difficult walking and balance task conditions. The concurrent changes of motor performance and cortical activation in reviewed studies contained the same direction change and different direction change.

Conclusion: Most of the reviewed studies reported poor motor performance and increased cortical activation of PD, stroke and multiple sclerosis older patients. The external motor performance such as step speed were analyzed only. The design and results were not comprehensive and profound. More than 5 weeks walking training or physiotherapy can contribute to motor function promotion as well as cortices activation of PD and stroke patients. Thus, further study is needed for more statistical analysis on the relationship between motor performance and activation of the motor-related cortex. More different type and program sports training intervention studies are needed to perform.

KEYWORDS

Parkinson, stroke, multiple sclerosis, gait, balance, postural control

1 Introduction

Neurological disorders encompass diseases of the brain and nervous system are the leading cause of disability (Murray et al., 2013) and contribute to 3% of disability-adjusted life years (Murray et al., 2012; Caliandro et al., 2020). Cerebrovascular injury (51%), neuromuscular disorders (7%), cognitive disorders (25%), and central nervous system infections (0.6%) are the common symptoms of older patients with neurological disorders (Bacellar et al., 2017), which would lead to dyskinesia (Defebvre and Krystkowiak, 2016; Harmon et al., 2019; Reich and Savitt, 2019). Dyskinesia would seriously affect older peoples' daily activities, especially walking and balance disability (Osoba et al., 2019). The expensive healthcare costs and additional neurology resource needs of longitudinal intervention burden the family and society. Parkinson's disease (PD), multiple sclerosis (MS), and stroke are common age-related neurological diseases interrelated with dyskinesia (Bonilauri et al., 2020).

The abnormal motor performance of stroke, PD, and MS patients, such as the step speed, step length, step width, step frequency, gait variability, stance time has been investigated (Hausdorff et al., 2007; Nutt et al., 2011; Socie and Sosnoff, 2013; Chisholm et al., 2014; Maidan et al., 2015; Belluscio et al., 2019). However, the investigation was not enough. Dyskinesia means impairment of control over ordinary muscle movement, which is not only associated with the degeneration or injury of the musculoskeletal or the nervous system but also associated with the complex linkage between them. Motor performance is one of the external representation of the musculoskeletal system, and brain cortical activity is one of the external representation of the central nervous system. If the changes and relationship of motor performance and cortical activity can be analyzed during the patients' movement, it would help probe the mechanism of dyskinesia and efficient rehabilitation methods for neurological disorder patients. However, testing the cortical activity of the brain during actual human movement is difficult.

The recent advancement in technologies such as Functional near-infrared spectroscopy (fNIRS), and portable electroencephalography has allowed for the investigation of brain function during realtime human movements in the natural environment freely. fNIRS is a non-invasive, repeatable, and reliable functional neuroimaging technology based on the theory of neurovascular coupling and optical spectroscopy (Villringer and Chance, 1997; Leff et al., 2011). An increase in neural activity of brain causes an increase in oxygen metabolism (Liao et al., 2013; Scholkmann et al., 2014; Pinti et al., 2020), leading to a decrease and increase in the concentration of oxygenated hemoglobin and deoxygenated hemoglobin (Lindauer et al., 2010; Liao et al., 2013; Scholkmann et al., 2014). The results of fNIRS has higher spatial resolution than portable electroencephalography, and has the highest correlation to functional magnetic resonance imaging BOLD measures (Strangman et al., 2002). In addition, fNIRS has been used to detect the cortical activity of the prefrontal cortex (PFC), primary cortex (M1), pre-motor cortex (PMC), supplementary motor area (SMA), and sensory-motor cortex (SMC) of the healthy or unhealthy population under single or dual-tasks during walking, turning, or balance intervention (Mihara et al., 2007; Al-Yahya et al., 2018; Stuart et al., 2018; Pelicioni et al., 2022), while portable electroencephalography based on neuroelectric signals of neurons is rarely used in dual-task gait activities compared with fNIRS. PFC takes part in planning, regulating and controlling of movement mainly (Szczepanski and Knight, 2014); SMC and M1 take part in planning, control and motor execution (Donoghue et al., 1994; Zwergal et al., 2012); SMA and PMC take part in planning and selecting movement (Gazzaniga, Ivry, and Mangun, 2009). Therefore, PFC, M1, PMC, SMA and SMC are all associated with movement.

Cortical activation and gait characteristics of PD patients (Stuart et al., 2019; Bonilauri et al., 2020), stroke patients (Chen et al., 2017; Herold et al., 2017), MS patients (Bonilauri et al., 2020), and cognitive impairment patients (Bishnoi et al., 2021) related to older neurological disorder patients with dyskinesia have been described under single or dual tasks during walking or balance intervention in former reviews touch on fNIRS. However, no review has tackled the relationship between motor performance and cortical activation of older neurological disorder patients while performing motor tasks. Therefore, this study aims to review the relationship between motor performance and cortical activity of older PD, stroke, and MS patients during walking and balance tasks. It might help improve patients' rehabilitation for intervention development to be uncovered.

2 Methods

2.1 Search strategy

Two independent researchers performed a systematic literature search in Scopus, PubMed, and Web of Science databases to identify all relevant studies published from 1 January 2012 to 30 December 2022 after the fNIRS became popular (Menant et al., 2020; Pinti et al., 2020). The 2020 Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) statements were used to report this systematic review. Key search terms and synonyms of three database are shown in Figure 1.

2.2 Eligibility criteria

Search and screening process is shown in Figure 2. Herein, studies were included based on the following criteria: 1) the sample or subgroup included age 60 years or older adults (Rudnicka et al., 2020) with neurological diseases such as PD, MS, and stroke, 2) fNIRS was used as the cortical activation test equipment, 3) gait or balance performance with kinematic results or score, and cortical activity were the primary result, 4) articles were written in English.

However, studies were deemed irrelevant based on the following criteria: 1) it was non-experimental studies (i.e., review and empirical studies), 2) participants were healthy people only, people with other diseases or animal, 3) other cortical activation test equipment (i.e., EEG), was used, 4) main results were not kinematics and cortical activation, 5) the full text of study was not available, 6) a brain-computer interface was used, 7) studies were the conferences, or case reports.

2.3 Study selection

A stepwise process was used to identify articles to be selected for inclusion in this review. Two authors independently screened

Search Terms Used for Each Database:	
Database	Search Terms
PubMed	("task" OR "motor" [Mesh] OR "motor") AND ("gait" [Mesh] OR "gait" OR "locomotion" [Mesh] OR "locomotion" OR "walking" [Mesh] OR "walking" OR "balance" [Mesh] OR "balance" OR "postural" [Mesh] OR "postural") AND ("elderly" OR "older") AND ("fNIRS" OR "functional near infrared*") AND (2012:2022[pdat])
Web of Science	((TS = (("walking" OR "gait" OR "locomotion" OR "motor" OR "balance" OR "postural") AND ("older"OR"elderly") AND ("fNIRS" OR "functional near infra *") AND ("task"))))
Scopus	("gait" [mesh] OR "locomotion" [mesh] OR "walking" [mesh] OR "walking" OR "balance" [mesh] OR "balance" OR "postural") AND ("elderly" OR "older" [mesh] OR "older") AND ("fNIRS" [mesh] OR "functional near infrared*") AND (LIMIT-TO (PUBYEAR,2022) OR LIMIT-TO (PUBYEAR,2021) OR LIMIT-TO (PUBYEAR,2020) OR LIMIT-TO (PUBYEAR,2019) OR LIMIT-TO (PUBYEAR,2018) OR LIMIT-TO (PUBYEAR,2017) OR LIMIT-TO (PUBYEAR,2016) OR LIMIT-TO (PUBYEAR,2015) OR LIMIT-TO (PUBYEAR,2014) OR LIMIT-TO (PUBYEAR,2013) OR LIMIT-TO (PUBYEAR,2012)) AND (LIMIT-TO (LANGUAGE,"English"))

FIGURE 1
Search string key terms used for study acquisition.

titles, followed by abstracts, and then full texts. Reviewers (YW, YD, WW, and YB) extracted the data and synthesized it into pre-designed tables. If any disagreement arose regarding the inclusion or exclusion of single studies, it would be resolved through consultation and confirmed by the reviewers (ZC and YT).

2.4 Data extract process

The following items were extracted from each included study; the name of author and published year; the type of neurological disorders (i.e., PD, stroke), age and degree of dyskinesia (i.e., freezing of gait); the number of fNIRS channels and region of interest; type of motor task and task paradigms (i.e., single task, dual task); intervention time; motor performance and cortical activation results including statistical correlation results, gait characteristics, balance score, and HbO_2 level of region of interest.

2.5 Risk of bias in individual studies

Each full-text article was assessed by two independent reviewers and scored using the PEDro scale. Any discrepancies were resolved by discussion and consensus. The PEDro scale was developed to measure methodological quality and internal validity of randomized studies (de Morton, 2009). Each of the 10 items is scored as either presenter absent (0) and a score is then calculated out of 10 with a larger number indicating better quality (Mason et al., 2016). The categories that were used to define overall quality of each article are as follows: ≦ 4 (poor), 5–6 (moderate) and ≧ 7 (high quality) (Fernandez et al., 2016).

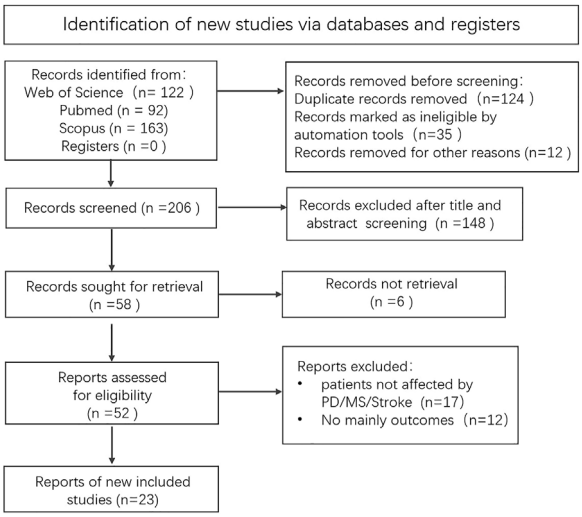

FIGURE 2
Search and screening process.

3 Results

3.1 Overview of included studies

As shown in Figure 2, a total of 377 studies were identified through key terms search, 206 studies were left in removed duplicates and so on, 148 were excluded in the title and abstract screening, 52 studies were reviewed in full texts, and 29 of them were excluded for not meeting the mentioned selection criteria. The remaining 23 studies were included in this review.

TABLE 1 Summary of motor performance and cortical activation outcomes in studies of Parkinson's patients.

Authors Year	Participant characteristics	fNIRS ROIs	Task paradigms	Motor performance and cortical activation results
Maidan et al. (2015)	11 PD (66.2 ± 10.0), FOG history	Channels (not mentioned)	ST1: normal walking	↓The turning acceleration, ↑HbO_2 level in PFC in FOG shows under ST2 condition compared to without FOG shows.
	11 HOA (71.2 ± 6.0)	PFC	ST2: 180° turn	
Maidan et al. (2016)	68 PD (71.7 ± 1.1)	6 channels	ST: normal walking	A positive correlation between HbO_2 level (↑) and step speed (↑) during obstacle negotiation walking. (*)
	38 HOA (70.4 ± 0.9)	PFC	DT1: ST + n-back	
			DT2: ST + obstacles negotiation	
Mahoney et al. (2016)	269 non-demented adults (76.41 ± 6.70):	16 channels	stand upright and count silently for 10 s	↑COP velocity, ↑ HbO_2 level in PFC in PS compared with HOA.
	117 MPS	PFC		
	26 PS			
	126 HOA			
Maidan et al. (2017)	49 PD (71.7 ± 1.0)	3 channels	2 min walk test (straight line + 180° turns)	A negative correlation between PFC activation (↑) and step speed (↓) during turning in the PD group. (*)
	126 HOA	PFC		
Al-Yahya et al. (2018)	29 PD (66.3 ± 5.9)	6 channels	ST: self-selected walking speed and fast walking speed	←Step characteristics, ↑HbO_2 level in PFC and M1 during DT condition compared with ST condition in PD group.
	22 HOA (59.5 ± 6.8)	PFC, M1	DT: ST + n-back	
Maidan et al. (2018)	64 PD:	6 channels	ST1: normal walking	↑Step speed and step length, ↓HbO_2 level in PFC during walking after treadmill training.
	34 TT (73.1 ± 1.1)	PFC	DT1: ST + n-back	
	30 TT + VR (70.1 ± 1.3)		DT2: ST + obstacles negotiation	
			6 weeks of treadmill training.	
Thumm et al. (2018)	20 PD (69.8 ± 6.5)	Channels (not mentioned)	ST1: overground walking	↓Step speed, ↓$Hb0_2$ level in PFC during ST2 condition compared to ST1 condition.
		PFC	ST2: treadmill walking	
Belluscio et al. (2019)	15 PD + FOG (66.9 ± 5.0)	8 channels	ST: 360° turns in place	A negative correlation between PFC activation (↑) and turning performance (↓) (↓number of turns in PD patients without FOG in DT during turning). (*)
	17PD + nonFOG (69.9 ± 4.3)	PFC	DT: ST + auditory AX-continuous	↓Step speed, ↓HbO_2 level in PFC for FOG during DT compared with ST.
	8 HOA (66.5 ± 5.5)			↓Step speed, ↑HbO_2 level during turning compared with HOA.
Stuart and Mancini (2020)	13 PD FOG (69.7 ± 4.2)	2 channels	ST1: 180° and 360° turns while walking	↓Step speed, ←PFC activity during walking or turning under DT condition compared with ST2 condition in the PD group.
	12 PD nonFOG (68.7 ± 3.9)	PFC	ST2: walking back and forth	
			DT: ST2+ auditoryAX-continuous with an open-or closed-loop tactile cue	
Vitorio et al. (2020)	24 PD + FoG (70.3 ± 4.7)	8 channels	ST: walking back and forth	↓Step speed and step length, ↓HbO_2 level in PFC under DT condition compared with ST condition in FOG group.
	23PD + nonFOG (70.8 ± 7.6)	PFC	DT: ST + auditoryAX-continuous	↑Step time variability, ↑HbO_2 level in PFC during walking in FOG group compared with nonFOG group.
Orcioli-Silva et al. (2021)	36 PD: 17 TD (70.8 ± 6.8) 19 PIGD (69.4 ± 6.6)	8 channels	ST: normal walking	↓Step speed and step length, ↑HbO_2 level in PFC during DT condition compared with ST condition.
		PFC	DT: obstacle negotiation walking	

(Continued on following page)

TABLE 1 (*Continued*) Summary of motor performance and cortical activation outcomes in studies of Parkinson's patients.

Authors Year	Participant characteristics	fNIRS ROIs	Task paradigms	Motor performance and cortical activation results
Hoang et al. (2022)	14 PD (67 ± 9)	8 channels DLPFC	ST: walking	A positive correlation between ΔHbO_2 of DLPFC (↓) and coefficient of variation of step length (↓). (*) ↑Step speed and step length, ↓HbO_2 level in DLPFC during T2 compared with T1.
			5 weeks of Sirocco training	
			T0 and T1: before training	
			T2: after training	
Pelicioni et al. (2022)	49 PD (69.5 ± 7.9)	6 channels	ST1: simple 60 s-walking trial	↓Step length and step speed, ↑PMC activation in PD group compared with HOA. ↓Step length and step speed, ↑SMA activation in ST2 condition compared with ST1 condition in PD group.
			ST2: obstacle negotiation	
	21 HOA (69.0 ± 5.9)	DLPFC, SMA, PMC	ST3: short and long target stepping	
			DT: ST1+ST2+ST3	

Abbreviations: HOA = healthy older adult, PD = Parkinson's disease, FOG = freezing of gait, PS = Parkinsonian syndromes, MPS = mild parkinsonian signs, TD = tremor dominant, PIGD = postural instability step disorder, Age = Mean ± SD, fNIRS = functional near infrared spectroscopy, PFC = prefrontal cortex, DLPFC = Dorsolateral prefrontal cortex, SMA = supplementary motor area, PMC = premotor cortex, ST = single-task, DT = dual-task, TT = treadmill training, VR = virtual reality, ↑:increase significantly, ↓:decrease significantly, ←: no significant differences, (*): Statistical correlation.

TABLE 2 Summary of motor performance and cortical activation outcomes in studies of stroke patients.

Authors Year	Participant characteristics	fNIRS ROIs	Task paradigms	Motor performance and cortical activation results
Mihara et al. (2012)	20 Stroke (61.9 ± 11.9)	50 channels	balance control task	A positive correlation between BBS score (↑) and the changes in HbO_2 (↑). (*)
		PFC, SMA		
Fujimoto et al. (2014)	20 Stroke (60.2 ± 9.5)	50 channels	balance control task	↑BBS score (postural stability) and ↑HbO_2 level in bilateral SMA after the intervention compared with before the intervention.
		PFC, SMA	about 6 weeks	
Mori et al. (2018)	14 Stroke (61.1 ± 9.3)	16 channels	ST: walking around a circle	↓Step speed and ↓HbO_2 level in PFC under DT condition in stroke group compared with HOA group.
	14 HOA (66.3 ± 13.3)	PFC	DT: ST + n-back	
Hawkins et al. (2018)	24 Stroke (58 ± 9.3)	4 channels	ST: normal walking	↓Step speed and ↑HbO_2 level in PFC under DT2 condition in stroke group compared with HOA group.
	15 HOA (77.2 ± 5.6)	PFC	DT1: ST + verbal letter fluency	
			DT2: ST + obstacles negotiation	
Caliandro et al. (2020)	22 Stroke (60 ± 8.1)	2 channels	walking with and without the wearable powered exoskeletons	A positive correlation between Knutsson score (↑) (abnormal muscle activation) of non-paretic limb and HbO_2 level in PFC (↑) during walking. (*)
	15 HOA (43–69)	PFC		
Collett et al. (2021)	45 Stroke (62 ± 14): 21 Good walker	32 channels	ST: normal walking	↓Step speed and step length, and↑HbO_2 level in PFC under DT condition, compare ST.
	24 Limited walker	PFC	DT: ST + stroop treadmill walking for approximately 10 weeks	
Lim et al. (2022a)	20 Stroke (64 ± 7.6)	54 channels	ST: normal walking	↓Step speed and ↑HbO_2 level in PFC, SMC, and PMC under DT condition compared with ST condition.
		PFC, SMC, PMC	DT: ST + verbal letter fluency	
Lim et al. (2022b)	20 Stroke (64 ± 7.6)	54 channels	normal walking	A positive correlation between step speed (↑) and cortical activity of PFC, SMC, and PMC (↑) of stroke under the acceleration stage of walking. (*)
		PFC, SMC, PMC		

Abbreviations: HOA = healthy older adult, Age = Mean ± SD, fNIRS = functional near infrared spectroscopy, PFC = prefrontal cortex, SMC = sensorimotor cortex, SMA = supplementary motor area, PMC = premotor cortex, ST = single-task., DT = dual-task, Berg Balance Scale = BBS, ↑: increase significantly, ↓: decrease significantly, (*): Statistical correlation.

TABLE 3 Summary of motor performance and cortical activation outcomes in studies of MS patients.

Authors Year	Participant characteristics	fNIRS ROIs	Task paradigms	Motor performance and cortical activation results
Hernandez et al. (2016)	8 MS (61 ± 4)	2 channels	ST: normal walking	↓Step speed and ↑HbO_2 level in PFC under DT condition compared with HOA or ST condition in MS group.
	8 HOA (57 ± 5)	PFC	DT: ST + reciting alternate letters	
Chaparro et al. (2017)	10 MS (56.2 ± 5.1)	16 channels	ST: walking on a treadmill with no or partial body weight support	↓Step speed and ↑HbO_2 level in PFC during walking compared with HOA.
	12 HOA (63.1 ± 4.4)	PFC	DT: ST + reciting alternate letters	

Abbreviations: HOA = healthy older adult, MS = multiple sclerosis, Age = Mean ± SD, fNIRS = functional near infrared spectroscopy, PFC = prefrontal cortex, ST = single-task, DT = dual-task, ↑: increase significantly, ↓: decrease significantly, (*): Statistical correlation.

TABLE 4 Hysiotherapy evidence database (PEDro) Scale scores of the reviewed studies.

PEDro criteria	1	2	3	4	5	6	7	8	9	10	Total
Maidan et al. (2015)	Yes	No	No	Yes	No	No	Yes	Yes	Yes	Yes	6
Maidan et al. (2016)	Yes	No	No	Yes	No	No	No	Yes	Yes	Yes	5
Mahoney et al. (2016)	Yes	No	No	Yes	No	No	Yes	Yes	Yes	Yes	6
Maidan et al. (2017)	Yes	No	Yes	Yes	No	No	No	Yes	Yes	Yes	6
Al-Yahya et al. (2018)	Yes	No	No	Yes	No	No	No	Yes	Yes	Yes	5
Maidan et al. (2018)	Yes	Yes	Yes	Yes	No	No	Yes	Yes	Yes	Yes	8
Thumm et al. (2018)	Yes	No	No	Yes	No	No	No	Yes	Yes	Yes	5
Belluscio et al. (2019)	Yes	No	No	Yes	No	No	No	Yes	Yes	Yes	5
Stuart and Mancini (2020)	Yes	No	No	Yes	No	No	No	Yes	Yes	Yes	5
Vitorio et al. (2020)	Yes	No	No	Yes	No	No	No	Yes	Yes	Yes	5
Orcioli-Silva et al. (2021)	Yes	No	No	Yes	No	No	No	Yes	Yes	Yes	5
Hoang et al. (2022)	Yes	No	No	Yes	No	No	No	Yes	Yes	Yes	5
Pelicioni et al. (2022)	Yes	Yes	No	Yes	No	No	No	Yes	Yes	Yes	5
Mihara et al. (2012)	Yes	Yes	No	Yes	No	No	No	Yes	No	Yes	5
Fujimoto et al. (2014)	Yes	Yes	No	Yes	No	No	No	Yes	No	Yes	5
Mori et al. (2018)	Yes	No	No	Yes	No	No	No	Yes	Yes	Yes	5
Hawkins et al. (2018)	Yes	No	No	Yes	No	No	No	Yes	Yes	Yes	5
Caliandro et al. (2020)	Yes	No	No	Yes	No	No	No	Yes	Yes	Yes	5
Collett et al. (2021)	Yes	No	Yes	Yes	Yes	No	No	Yes	Yes	Yes	7
Lim et al. (2022a)	Yes	No	No	Yes	No	No	No	Yes	Yes	Yes	5
Lim et al. (2022b)	Yes	Yes	No	Yes	No	No	No	Yes	No	Yes	5
Hernandez et al. (2016)	Yes	No	No	Yes	No	No	Yes	Yes	Yes	Yes	6
Chaparro et al. (2017)	Yes	No	No	Yes	No	No	Yes	Yes	Yes	Yes	6

Criteira: 1) Eligibility; 2) Randomized allocation; 3) Concealed allocation; 4) Baseline similarity; 5) Blinded subjects; 6) Blinded therapists; 7) Blinded assessors; 8) Key outcomes; 9) Comparison between groups; 10) Point measures and measures of variability.

Tables 1–3 showed the 23 studies included in this study. Of these, studies assessed patients with PD (N = 13), with a previous history of stroke (N = 8), and with MS (N = 2). The age of older patients in these studies was widely distributed from 56 ± 5 to 73 ± 1 year. Studies contain cross-sectional (N = 19) and longitudinal designs (N = 4).

Among the 23 studies, walking was used as a test task in 20 studies, including straight walking (N = 15), turning (N = 5), obstacles (N = 5), and treadmill walking (N = 4). Meanwhile, the balance was used as a test task in the other three studies, including normal standing (N = 1) and postural intervention standing (N = 2).

Gait and posture stability were the primary behavioral outcomes of walking and balance tasks. PFC was the most popular region of interest (N = 19) which was tested by 2–32 channels in fNIRS systems. Other region of interest in the frontal and parietal cortices were tested by six channels (N = 2) or more than 40 channels in the fNIRS system (N = 4).

3.2 Risk of bias in individual studies

Scores on the PEDro scale ranged from five to eight for the selected articles (Table 4). Twenty one studies were of moderate quality; and two studies were high quality.

3.3 Co-direction change of motor performance and cortical activation

Seven studies reported increased gait or balance results and increased cortical activation. One of them found a positive correlation between step speed and cortical activation of PFC during obstacle negotiation walking of PD patients *via* Pearson's correlation methods (Maidan et al., 2016). Three of them found a positive correlation between Berg Balance Scale, and cortical activation of PFC and SMA of stroke patients *via* Spearman's correlation methods (Mihara et al., 2012), Knutsson score (abnormal muscle activation) and cortical activation of PFC of stroke patients with exoskeletons by multiple regression methods (Caliandro et al., 2020); and step speed and cortical activity of PFC, SMC, and PMC of stroke patients *via* Pearson's correlation methods (Lim et al., 2022b). Three studies found that Center of Pressure velocity and cortical activation of PFC were increased (Mahoney et al., 2016); step time variability and cortical activation of PFC of PD patients with freezing of gait compared with non-freezing of gait during walking were increased (Vitorio et al., 2020); BBS score and cortical activation of SMA were increased during balance task after neuro-modulative therapies rehabilitation (Fujimoto et al., 2014).

Five studies reported decreased gait or balance results and decreased cortical activation. One of them found a positive correlation between coefficient of variation of step length, and cortical activation of dorsolateral prefrontal cortex of PD patients *via* Pearson's correlation methods (Hoang et al., 2022). Three of them found step speed and PFC cortical activation of PD patients were decreased during dual-task condition compared with single-task condition (Belluscio et al., 2019; Vitorio et al., 2020), and during treadmill walking compared with over-ground walking (Thumm et al., 2018). One of them found step speed and cortical activation of PFC were decreased under dual-task condition of stroke patients compared with healthy older adults (Mori., 2018).

3.4 Inverse change of motor performance and cortical activity

Ten studies reported decreased gait or balance results and increased cortical activation. Two of them found negative correlations between increased PFC activation and decreased step speed (Maidan et al., 2017), and the decreased number of turns (Belluscio et al., 2019) of PD patients under dual-task conditions *via* Pearson's correlation methods. Two of them found that step speed was decreased (Belluscio et al., 2019; Pelicioni et al., 2022), cortical activation was increased in PFC (Belluscio et al., 2019) and PMC (Pelicioni et al., 2022) of PD patients compared with healthy participants. One of them found that turning acceleration was decreased during turning, and cortical activation of PFC was increased in PD patients with freezing of gait compared with non-freezing of gait (Maidan et al., 2015). Two of them found that step speed and length were decreased, cortical activation of PFC (Orcioli-Silva et al., 2021) and SMA (Pelicioni et al., 2022) were increased under obstacle negotiation dual-task conditions compared with single task condition. Three of them found step speed was decreased, cortical activation of PFC (Hawkins et al., 2018; Collett et al., 2021; Lim et al., 2022a), SMC (Lim et al., 2022a), and PMC (Lim et al., 2022a) were increased in stroke patients during walking under dual-task conditions compared with single-task condition (Collett et al., 2021; Lim et al., 2022a) or compared with healthy older adults (Hawkins et al., 2018). Two studies of MS patients found that step speed was decreased and cortical activation of PFC was increased under dual-task conditions (Hernandez et al., 2016) during walking compared to healthy older adults (Chaparro et al., 2017).

Two longitudinal studies reported step speed and length were increased, cortical activation of PFC or dorsolateral PFC was decreased after intervention (Maidan et al., 2018; Hoang et al., 2022).

One study reported gait characteristics did not change while cortical activation of PFC and M1 were increased of PD patients under dual-task conditions compared with single task condition or healthy older adults (Al-Yahya et al., 2018); the other one study reported step speed was decreased while cortical activation of PFC did not change of PD patients under dual-task conditions compared with single-task condition (Stuart and Mancini, 2020).

4 Discussion

A total of 23 studies that investigated the motor performance and cortical activity *via* fNIRS were systematically reviewed in this study. It showed varied relationships between motor performance and cortical activity of older adults with PD, stroke, and MS patients under walking and balance tasks.

4.1 Poor performance with changed cortical activation

Among the 23 studies, it was found that step speed and length as well as the number of turns were decreased, and the Center of Pressure velocity, step time variability, Knutsson score were increased. Those results usually used to indicate poor motor performance (Mancini et al., 2017; Belluscio et al., 2019; Kahya et al., 2019; Caliandro et al., 2020). However, the cortical activation of PFC, PMC, SMC, and SMA was increased or decreased during different motor and balance task.

Poor motor performance accompanied by increased PFC, PMC, SMC, and SMA activation indicated that automatic walking function was affected by the disease (Herold et al., 2017; Maidan

et al., 2017; Pelicioni et al., 2022), because walking was controlled by two pathways in the brain (la Fougère et al., 2010; Zwergal et al., 2012; Zwergal et al., 2013; Herold et al., 2017). The direct and normal pathways relevant to automatic movement are controlled by the striatum and activated in the case of low-challenging walking and the absence of pathology. The indirect pathway relevant to motor-related cognitive resources involving the PFC, PMC, and SMC areas is activated in the impaired automatic walking caused by aging or disease (Vandenbossche et al., 2012; Clark, 2015; Herold et al., 2017; Al-Yahya et al., 2018). SMA cortex was also involved in inter-limb coordination, gait, and postural control (Fujimoto et al., 2014), which might activate in the complex obstacle negotiation task.

Therefore, poor motor performance with increased cortical activation might indicate that automatic walking turns into voluntary cortical control walking (Maidan et al., 2015), which might be a cognition and postural compensation to counter the motor or neuron dysfunction of neurological disorder patients with dyskinesia.

However, poor motor performance accompanied by decreased PFC activation indicates that once the compensatory is over-activated, the cognitive resources would reach the limit (Holtzer and Izzetoglu, 2020). Dual-task and circle walking are both difficult tasks for neurological disorder patients with dyskinesia. The changes in brain structure and function caused by that disease might easily induce the inflexibility or overburdening of executive attention cognitive resources (Stuart and Mancini, 2020). Consequently, the neural efficiency of PFC becomes lower, the executive function goes down, the cortical activation of PFC reaches the ceiling, the impaired motor function cannot get support from the cortex, and the motor performance becomes poor.

4.2 Better performance with changed cortical activation

Among the 23 studies, it was found that step speed, BBS score and length were increased, and the coefficient of variation of step length and step speed were decreased. Those results usually used to indicate improved motor performance (Collett et al., 2021; Hoang et al., 2022), and improved balance control (Fujimoto et al., 2014). However, sometimes the cortical activation of PFC decreased, sometimes the cortical activation of PFC, SMA, PMC, and SMC increased during different motor task.

Better motor performance accompanied by decreased PFC activation showed a positive effect of sports training on neurological disorder patients with dyskinesia. It indicates that the automaticity of walking was improved, the dependence on cognitive resources was reduced, and the patients could make better use of their executive resources to walk normally (Hoang et al., 2022). It might be because sports training enhances neural plasticity to a certain extent, thereby promoting angiogenesis, neurogenesis, and synaptogenesis, all of which led to an increased efficiency and help reduce the required levels of cortex activation (Maidan et al., 2018).

Better motor performance accompanied by increased PFC, PMC, SMA, and SMC activation also showed positive effects on balance and normal walking tasks. Based on the compensatory reallocation model (Verghese et al., 2017), the brain recruits neural-related area and allocates more neural resources to the motor program when the task was slightly difficult for patients, thereby resulting in near-normal and safe motor performance finally. Therefore, higher cortical activation caused by balance and walking helped patients with limited motor ability to improve their motor performance.

Better motor performance accompanied by increased SMA activation showed a positive effect on balance task skills. SMA was related to inter-limb coordination and postural control (Fujimoto et al., 2014), and was a potential target of neuro-modulative therapies which can promote recovery of motor function (Hummel and Cohen, 2006). Therefore, the advanced SMA cortical activation would help recover the balance function of stroke patients.

4.3 Changed performance or cortical activation *versus* unchanged

Among the 23 studies, one study reported unchanged step character accompanied by increased cortical activation of PFC and M1, the other one reported decreased step speed accompanied by unchanged cortical activation of PFC under dual-task conditions during walking.

Based on the compensatory reallocation model, in some motor tasks, the brain recruits neural network tissue in the motor-related cortex and allocates more neural resources to the motor program, thereby resulting in near-normal motor performance and high cortical activation (Verghese et al., 2017). Therefore, the brain would regulate the cortex resources and help patients to complete difficult walking. Patients with limited motor ability can maintain motor performance through improved cortical activation. However, if the resources of PFC are occupied by the additional task and reach the ceiling, it might be hard for patients to maintain motor performance well (Holtzer and Izzetoglu, 2020).

4.4 Learning from the change of relationships

Human movement is executed by the muscles and controlled by the Nervous system (Clark et al., 2014). The muscle fibers receive the input information from alpha neurons. The alpha neurons receive the input information from spinal fibers and neurons. The spinal receives input from extra-pyramidal tracts. The extra-pyramidal tracts receive the input from the cortical (i.e., PFC, M1, PMC, and SMA) and subcortical structure (Gazzaniga et al., 2009). Therefore one of symptom of neurological disorder patient is dyskinesia. The dyskinesia is associated with the abnormal of motor control which showed abnormal cortical activation and motor performance.

With the development of the realtime human movement cortical activation test equipment, more and more studies follow the interest of abnormal motor control of dyskinesia neurological disorder patients. However, the design and results were not comprehensive and profound. Most of the reviewed studies reported poor motor performance and increased cortical activation of PD, stroke and MS older patients under simple motor task conditions, compare with healthy older adults or

difficult motor task conditions. Few studies discussed the exact relationship between motor performance and cortical activation, or used statistical analysis. The external motor performance such as step speed and step length were analyzed only. The sports biomechanics variables (i.e., muscle activation, segment movement) and its relationship with cortical activation were left behind.

Our review results showed that motor performance has improved after sports training or occupational therapy intervention. However, the type of sports training was simple, the difficulty, intensity, time interval of sports training intervention needs further study. The expanded and specific study would promote the clinical rehabilitation intervention therapy of neurological disorder patient with dyskinesia.

5 Conclusion

From the 23 included publications, four were interventional studies, seven studies used statistical methods to interpret the relationship between motor performance and cortical activation, participants were mostly PD patients, PFC was the favorite region of interest, and walking was the most popular test motor task. The motor performance and cortical activation of frontal and parietal cortices were simultaneously affected during difficult walking and balance tasks of the neurological disorder patients with dyskinesia. Most reviewed studies reported poor motor performance and increased cortical activation of PD, stroke and MS older patients, compare with healthy older adults or difficult motor task conditions. Few studies discussed the exact relationship of motor performance and cortical activation by statistical analysis. The design and results were not comprehensive and profound. The sports biomechanics variables and its relationship with cortical activation variables were left behind. More than 5 weeks of walking training or physiotherapy can contribute to motor function promotion as well as frontal and parietal cortices activation of PD and stroke patients.

6 Limitation

There are several limitations in the present review. Firstly, it was limited to studies published in English. As such, studies with relevant findings may have been ommited. Secondly, the search time period for this review was the recent 10 years while fNIRS became popular. As such, we might not identify all relevant articles. Thirdly, it was limited to studies containing walking and balance motor tasks. As such, studies relevant to upper extremity motor tasks were excluded. Fourthly, the other cortical activation detected methods, such as portable electroencephalography were not included in this review, it is not known if a different detected method could have different results. Last but not least, the various study design made it difficult to summarize and discuss the different neural pathway changes of walking and balance control.

7 Future scope

Further study is needed to do more statistical analysis on the relationship between motor performance and activation of motor-related cortex. It would help analyze their relationship accurately. And dual task walking, obstacle walking and turning might be used as a sports training method. Intervention studies should be conducted to distinguish the intervention time, intensity and effects of dual-task walking, obstacle walking, over ground walking, and treadmill walking for neurological disorder patients with dyskinesia. It would help to look for new sport rehabilitation methods for those patients. Meanwhile, body performance such as trunk or head movement, electromyography, and ground reaction force should be synchronized with cortical activation test to evaluate the posture control of patients during walking and balance task. It would help to investigate the common change in musculoskeletal and nervous system, then further discuss their collaboration. Finally, the relationship between the different neural pathways of movement and the corresponding motor performance should be considered. It would help to investigate the different coordinate mechanisms of the musculoskeletal and nervous system in diverse patients movements.

Data availability statement

The original contributions presented in the study are included in the article/supplementary material, further inquiries can be directed to the corresponding author.

Author contributions

YW and YD gathered and checked the references, wrote each draft, and formatted the manuscript to meet journal requirements. YT, WW, and YB led the research topic and commented on and added to each draft. CZ developed the concept for the paper, checked the references, wrote each draft, and commented on and added to each draft.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Surface electromyography characteristics of patients with anterior cruciate ligament injury in different rehabilitation phases

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Background: Anterior cruciate ligament reconstruction (ACLR) is a common treatment for anterior cruciate ligament (ACL) injury. However, after ACLR, a significant proportion of patients do not return to pre-injury levels. Research on muscle function during movement has important implications in rehabilitation.

Methods: Sixty patients with unilateral ACL injury were recruited for this study and assigned into three groups: group A, individuals with an ACL injury before 6 months; group B, individuals with ACLR from 6 months to 1 year; and group C, individuals with ACLR 1 year later. Surface electromyography (SEMG) signals were collected from the bilateral rectus femoris (RF), vastus medialis (VM), vastus lateralis (VL), biceps femoris (BF), and semitendinosus (ST). The tasks performed during the experiment included straight leg raising (SLR) training at 30°, SLR training at 60°, ankle dorsiflexion, walking, and fast walking.

Results: In the maximum muscle strength test, the affected side of the BF in group A (199.4 ± 177.12) was significantly larger than in group B (53.91 ± 36.61, $p = 0.02$) and group C (75.08 ± 59.7, $p = 0.023$). In the walking test, the contralateral side of the RF in group B (347.53 ± 518.88) was significantly greater than that in group C (139.28 ± 173.78, $p = 0.029$). In the SLR training (60°) test, the contralateral side of the RF in group C (165.37 ± 183.06) was significantly larger than that in group A (115.09 ± 62.47, $p = 0.023$) and smaller than that in group B (226.21 ± 237.17, $p = 0.046$); In the ankle dorsiflexion training test, the contralateral side of the RF in group B (80.37 ± 87.9) was significantly larger than that in group C (45.61 ± 37.93, $p = 0.046$).

Conclusion: This study showed the EMG characteristics of patients with ACL injury helped to determine which muscle requires more training and which exercise model would be best suited for intervention.

KEYWORDS

anterior cruciate ligament injury, surface electromyography, rehabilitation, muscle function, muscle strength

1 Introduction

Anterior cruciate ligament reconstruction (ACLR) is a common treatment option for patients with anterior cruciate ligament (ACL) injuries (Van Melick et al., 2016). It is estimated that up to 200,000 ACLRs are performed annually in the United States (Fryer et al., 2019). Patient success rate for primary ACLR is approximately 75%–97% (Kim et al., 2018). However, after ACLR, a significant proportion of patients do not return to their pre-injury levels. A systematic review and meta-analysis by Ardern et al. showed that 69 studies found 81% of individuals reported returning to sports after ACLR. In contrast, only 65% reported returning to the pre-injury level of sports participation (Ardern et al., 2014). Recent research shows that 35% of athletes do not return to the pre-injury sports level within 2 years after ACLR, and half reported their ACL injury as the primary reason for a lower activity level (Ardern et al., 2011; Ardern et al., 2012; Ardern et al., 2014). The prevalence of these adverse outcomes underscores the importance of rehabilitation after ACLR, so further research is warranted on how to improve the effectiveness of rehabilitation.

In addition to restoring joint stability, ACLR surgery aims to restore knee joint function and muscle strength (Kim et al., 2015). An appropriate rehabilitation program was incorporated into routine preoperative and postoperative care to maximize surgical outcomes and improve functional recovery. Impaired knee function associated with ACL injury includes instability in dynamic movements and weakness of the quadriceps (Kim et al., 2018). Quadriceps muscle weakness, critical to dynamic joint stability, eventually leads to decreased knee function and poor exercise performance and can contribute to the early onset of osteoarthritis (Muaidi et al., 2007). Thus, the function of the muscles around the knee joint, especially the quadriceps femoris (QF), is the main training target and serves as an indicator for monitoring functional recovery after ACLR surgery (Slemenda et al., 1997). After ACLR, most patients have lower extremity muscle strength deficits on the affected and contralateral sides. Specifically, between-limb QF muscle strength symmetry is recommended as an important clinical benchmark to determine whether an athlete is ready for sports after ACLR (Logerstedt et al., 2017).

FIGURE 1
Electrode positions are shown in the Figure.

An increasing number of studies have focused on evaluating the primary ACLR surgery method, joint stability, muscle strength, and knee function (Muaidi et al., 2007; Kim et al., 2015). Jong et al. reported that lower extremity muscle atrophy and weakness after ACLR represent a difficult and unresolved problem (De Jong et al., 2007). Unresolved post-operative muscle strength deficits might be associated with knee osteoarthritis that is present years after surgery (Keays et al., 2010). Investigators have also reported post-operative strength deficits ranging from 5% to 40% for the quadriceps and from 9% to 27% for the hamstrings (De Jong et al., 2007; Hiemstra et al., 2007; Karanikas et al., 2009). However, limited studies have assessed muscle strength recovery after revision ACLR surgery during different sports modes.

Limited recovery of lower extremity muscle function may be related to the current rehabilitation program. Most studies have described current time-based rehabilitation protocols that are mainly based on the remodeling process of the graft (Czuppon et al., 2014). However, there is still uncertainty regarding the schedule of the human remodeling process, and more experts are considering incorporating functional goal-based criteria into the rehabilitation protocol (Howells et al., 2011). In addition, there were individual differences in neuromotor learning and flexibility after ACLR. These findings underscore the importance of a change from time-based rehabilitation to goal-based rehabilitation with neuromuscular goals and criteria to manage the rehabilitation process. Österberg et al. (2013) suggested that these goals for progression to the next phase and description of interventions during each phase should be based on the International Classification of Functioning, Disability, and Health, which may be more suitable for the rehabilitation of patients. Among these studies on rehabilitation after ACLR, few studies have focused on the evaluation of quality of movement. However, the relevance of focusing more on the quality of movement is underlined by the fact that altered neuromuscular function and biomechanics after ACLR could be risk factors for a second ACL injury (Hiemstra et al., 2009).

Surface electromyography (SEMG) has the advantages of non-invasion, real-time, and multitarget measurement, and is a method that has received increasing attention due to its ability to quantitatively analyze neuromuscular activity in static and dynamic motion states (Rasool et al., 2017). SEMG has been used to assess normal and abnormal muscle activation in some patients with ACL injury to guide rehabilitation strategies (Zebis et al., 2019). Nevertheless, it is rarely used to assess muscle activation at different stages of rehabilitation.

After ACL injury and reconstruction, lower extremity muscle strength has been reported to decrease not only in the quadriceps but also in the hamstrings, usually continuing after the postoperative rehabilitation period. In vigorous dynamic movements, coactivation

TABLE 1 SEMG values of maximum muscle strength test between different groups.

	Groups	Mean ± SD	Groups		P-value
Affected side -RF	A	329.98 ± 236.46	A	B	0.515
	B	210.03 ± 110.6	A	C	0.924
	C	314.63 ± 402.42	B	C	0.518
Contralateral side -RF	A	400.3 ± 326.11	A	B	0.201
	B	155.27 ± 122.3	A	C	0.916
	C	417.73 ± 381.76	B	C	0.123
Affected side -VL	A	531.33 ± 653.88	A	B	0.074
	B	134.77 ± 58.04	A	C	0.176
	C	272.36 ± 223.1	B	C	0.465
Contralateral side -VL	A	329.33 ± 130.88	A	B	0.104
	B	121.21 ± 92.71	A	C	0.674
	C	375.18 ± 277.15	B	C	**0.028**
Affected side -VM	A	182.3 ± 163.28	A	B	0.347
	B	122.5 ± 84.99	A	C	0.798
	C	168.13 ± 78.7	B	C	0.413
Contralateral side -VM	A	353.63 ± 299.69	A	B	0.053
	B	127.25 ± 81.33	A	C	0.312
	C	253.36 ± 155.73	B	C	0.207
Affected side -BF	A	199.4 ± 177.12	A	B	**0.02**
	B	53.91 ± 36.61	A	C	**0.023**
	C	75.08 ± 59.7	B	C	0.681
Contralateral side -BF	A	205.18 ± 201.7	A	B	0.91
	B	184.1 ± 260.93	A	C	0.825
	C	169.05 ± 384.54	B	C	0.927
Affected side -ST	A	446.4 ± 435.73	A	B	**0.014**
	B	93.25 ± 91.91	A	C	**0.003**
	C	59.07 ± 61.64	B	C	0.77
Contralateral side -ST	A	260.42 ± 243.44	A	B	0.252
	B	53.05 ± 49.4	A	C	0.613
	C	181.01 ± 393.31	B	C	0.418

SEMG, surface electromyography; RF, rectus femoris; VM, vastus medialis; VL, vastus lateralis; BF, biceps femoris; ST, semitendinosus.
That the bold values indicates the statistical significance was established.

of the hamstring tendons is important to provide dynamic stability of the knee and prevent excessive shear forces of the ACL. Kim et al. found that after an ACL injury and subsequent reconstruction, deficits in quadriceps muscle strength were significant (Kim et al., 2015). Bade et al. showed that in early high intensity and low intensity rehabilitation after total knee, quadriceps and hamstring strength and quadriceps activation improved beyond baseline performance in both groups (Bade et al., 2017). Thus, we performed this study to assess lower muscle functional recovery before and after ACLR surgery at different periods using different sports modes. We aimed to 1) compare the recovery of knee extensor muscle function (rectus femoris [RF], vastus medialis [VM], vastus lateralis [VL]) and flexor muscle (biceps femoris [BF] and semitendinosus [ST]) before and after ACLR and 2) compare the functional outcomes of different sports modes. We hypothesized that patients with ACL injury would exhibit different characteristics of SEMG in different rehabilitation phases and in different training tasks.

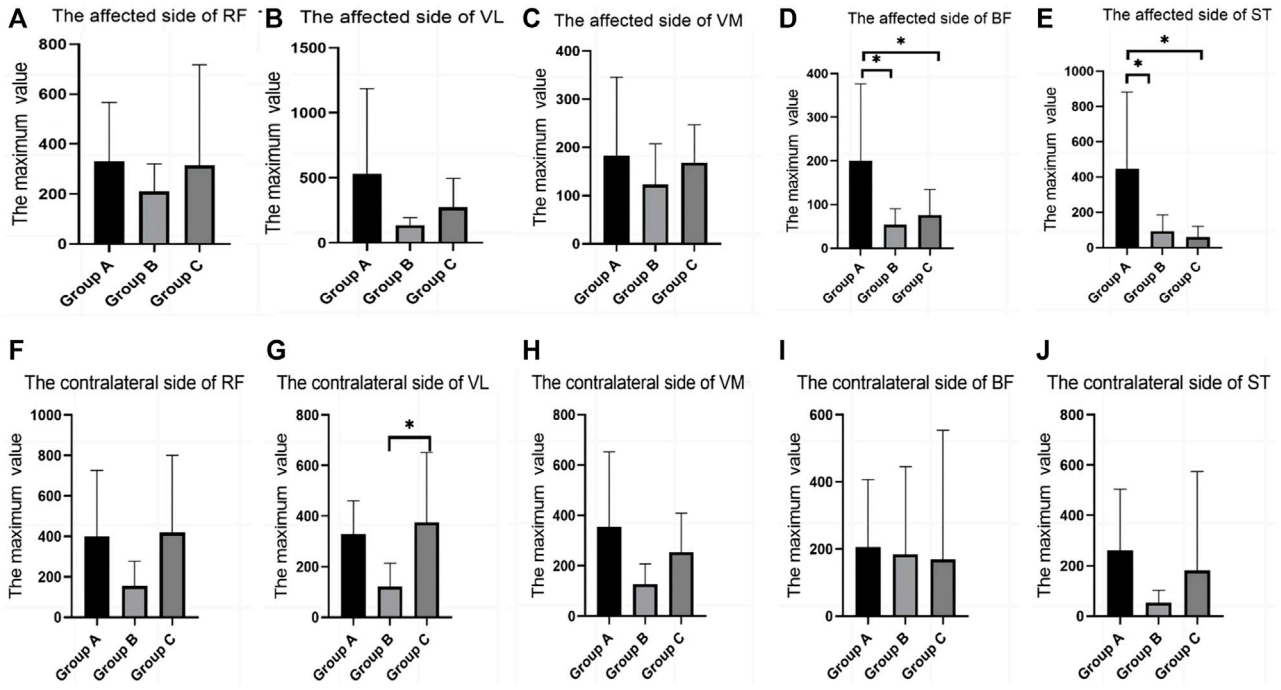

FIGURE 2
SEMG values of maximum muscle strength test between different groups. **(A)**, The affected side of RF; **(B)**, The affected side of VL; **(C)**, The affected side of VM; **(D)**, The affected side of BF; **(E)**, The affected side of ST; **(F)**, The contralateral side of RF; **(G)**, The contralateral side of VL; **(H)**, The contralateral side of VM; **(I)**, The contralateral side of BF; **(J)**, The contralateral side of ST; SEMG, surface electromyography; RF, rectus femoris; VM, vastus medialis; VL, vastus lateralis; BF, biceps femoris; ST, semitendinosus.

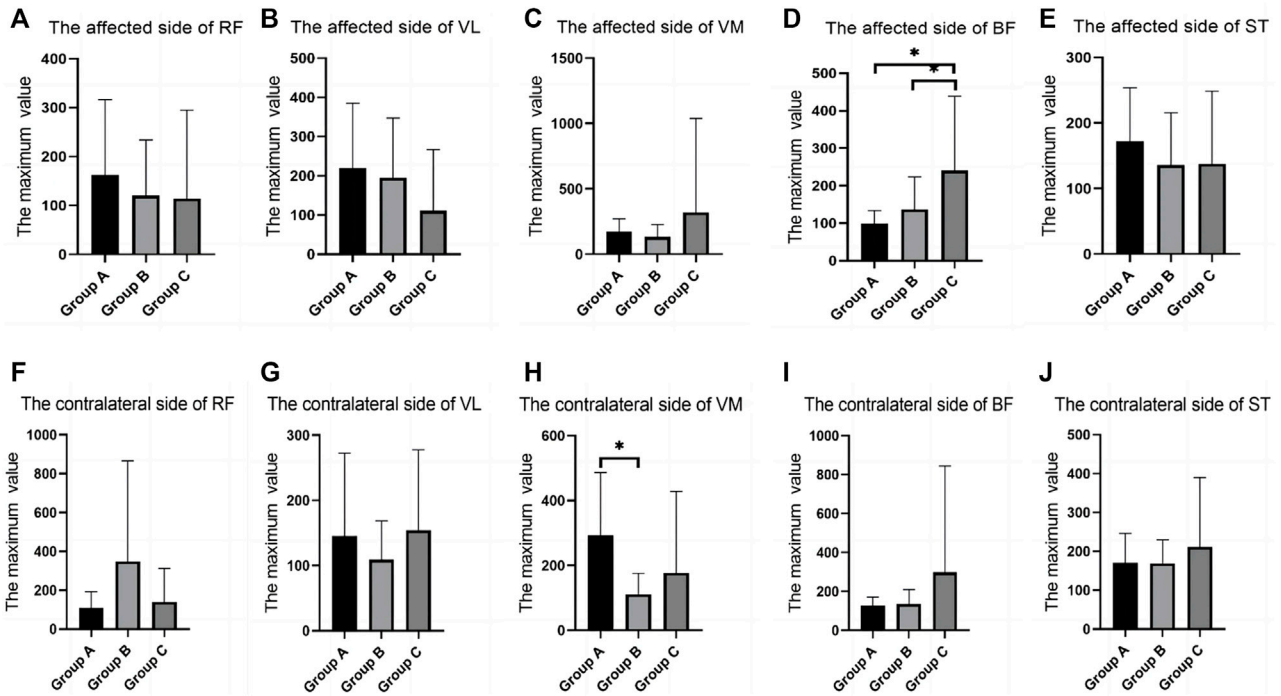

FIGURE 3
SEMG values of different groups during walking. **(A)**, The affected side of RF; **(B)**, The affected side of VL; **(C)**, The affected side of VM; **(D)**, The affected side of BF; **(E)**, The affected side of ST; **(F)**, The contralateral side of RF; **(G)**, The contralateral side of VL; **(H)**, The contralateral side of VM; **(I)**, The contralateral side of BF; **(J)**, The contralateral side of ST; SEMG, surface electromyography; RF, rectus femoris; VM, vastus medialis; VL, vastus lateralis; BF, biceps femoris; ST, semitendinosus.

TABLE 2 SEMG values of different groups during walking.

	Groups	Mean ± SD	Groups		P-value
Affected side -RF	A	162.16 ± 154.37	A	B	0.527
	B	120.09 ± 114.12	A	C	0.416
	C	114.09 ± 181.05	B	C	0.906
Contralateral side -RF	A	109.58 ± 82.28	A	B	0.055
	B	347.53 ± 518.88	A	C	0.784
	C	139.28 ± 173.78	B	C	**0.029**
Affected Side -VL	A	219.27 ± 166.07	A	B	0.705
	B	194.95 ± 152.2	A	C	0.062
	C	111.09 ± 155.7	B	C	0.093
Contralateral side -VL	A	145.72 ± 126.89	A	B	0.423
	B	109.09 ± 59.27	A	C	0.833
	C	154.26 ± 123.45	B	C	0.199
Affected side -VM	A	173.25 ± 96.18	A	B	0.853
	B	131.67 ± 95.65	A	C	0.458
	C	321.24 ± 717.19	B	C	0.273
Contralateral Side -VM	A	292.86 ± 192.94	A	B	**0.038**
	B	111.61 ± 64.62	A	C	0.133
	C	177.78 ± 250.88	B	C	0.315
Affected Side -BF	A	99.45 ± 34.28	A	B	0.561
	B	137.07 ± 86.54	A	C	**0.017**
	C	240.03 ± 198.78	B	C	**0.042**
Contralateral side -BF	A	127.68 ± 43.97	A	B	0.958
	B	136.57 ± 72.78	A	C	0.264
	C	297.65 ± 546.23	B	C	0.221
Affected side -ST	A	172.25 ± 81.63	A	B	0.368
	B	135.52 ± 80.08	A	C	0.34
	C	137.63 ± 110.93	B	C	0.946
Contralateral side -ST	A	169.74 ± 76.27	A	B	0.997
	B	169.55 ± 60.07	A	C	0.423
	C	211.25 ± 178.38	B	C	0.352

SEMG, surface electromyography; RF, rectus femoris; VM, vastus medialis; VL, vastus lateralis; BF, biceps femoris; ST, semitendinosus.
That the bold values indicates the statistical significance was established.

2 Methods

2.1 Recruitment of participants

In this study, based on literature reports and previous research data, using G*Power software (3.1.9.7) calculated sample size, sixty patients with unilateral ACL injuries were recruited. Patients who met the following criteria were included: 1) age >18 years; 2) diagnosed as ACL injury by magnetic resonance imaging and clinical examination or had received ACLR surgery using the all-soft tissue quadriceps tendon. The excluded criteria were the following: 1) patients with previous ACLR surgery and recurrence of ACL injury. 2) patients with cognitive impairment unable to complete experimental procedures; 3) patients with a history of other neurological diseases or disorders, lower extremity surgery, or fracture. Patients eligible for enrolment were divided into three groups according to the time of injury and the time to surgery. All patients completed routine preoperative rehabilitation. Individuals with ACL injury less than half a year were recruited as Group A (16 men and 4 women, average age: 33.5 ± 11.77 years;

TABLE 3 SEMG values during fast walking.

	Groups	Mean ± SD	Groups		P-value
Affected side -RF	A	241.22 ± 274.33	A	B	0.993
	B	240.71 ± 188.02	A	C	0.079
	C	141.08 ± 99.02	B	C	**0.044**
Contralateral side -RF	A	391.95 ± 430.19	A	B	0.557
	B	500.38 ± 828.46	A	C	0.257
	C	199.96 ± 227.7	B	C	**0.042**
Affected side -VL	A	306.77 ± 324.57	A	B	0.324
	B	248.57 ± 128.09	A	C	**0.007**
	C	158.12 ± 67.57	B	C	0.055
Contralateral side -VL	A	171.33 ± 86.88	A	B	0.281
	B	210.73 ± 92.93	A	C	0.233
	C	211.24 ± 111.1	B	C	0.986
Affected side -VM	A	276.23 ± 154.71	A	B	0.825
	B	235.61 ± 214.4	A	C	0.44
	C	406.61 ± 692.31	B	C	0.243
Contralateral side -VM	A	362.23 ± 182.44	A	B	0.31
	B	217.02 ± 195.37	A	C	0.783
	C	326.33 ± 522.64	B	C	0.333
Affected side -BF	A	245.77 ± 143.18	A	B	0.661
	B	280.05 ± 282.65	A	C	0.756
	C	267.97 ± 198	B	C	0.845
Contralateral side -BF	A	215.38 ± 125.05	A	B	0.924
	B	215.38 ± 196.95	A	C	0.24
	C	371.69 ± 537.51	B	C	0.175
Affected side -ST	A	171.4 ± 81.42	A	B	0.108
	B	341.97 ± 465.56	A	C	0.417
	C	249.48 ± 199.2	B	C	0.267
Contralateral side -ST	A	222.96 ± 111.68	A	B	0.547
	B	262.21 ± 203.31	A	C	0.945
	C	218.91 ± 186.31	B	C	0.404

SEMG, surface electromyography; RF, rectus femoris; VM, vastus medialis; VL, vastus lateralis; BF, biceps femoris; ST, semitendinosus.
That the bold values indicates the statistical significance was established.

height: 171 ± 5.36 cm; weight: 70.1 ± 13.71 kg); Individuals with ACLR for less than half a year were recruited as Group B (13 men and 7 women, average age: 32.35 ± 9.4 years; height: 171.85 ± 8.46 cm; weight: 75.98 ± 17.49 kg); Individuals with ACLR more than half a year and less than 1 year were recruited as Group C (14 men and 6 women, average age: 31.45 ± 8.02 years; height: 171.35 ± 7.81 cm; weight: 71.88 ± 12.25 kg). Patients with knee disorders, anatomical abnormalities, or a history of surgery were excluded. The Medical Ethics Committee of Binzhou Medical University Hospital approved this study (2021-S019-01). A signed consent form was obtained from each participant before the test was performed.

2.2 Measurement process

Surface electromyography (SEMG) was performed using a Noraxon wireless dynamic electromyography tester (Noraxon,

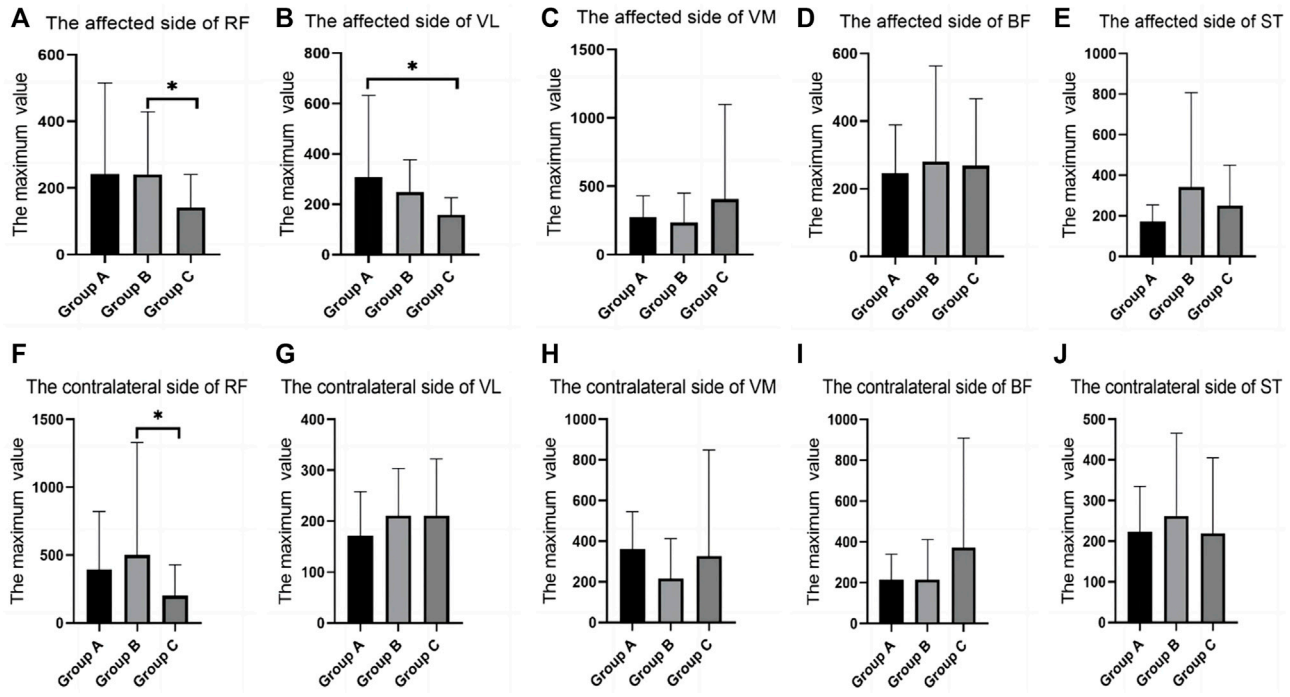

FIGURE 4
SEMG values during fast walking. **(A)**, The affected side of RF; **(B)**, The affected side of VL; **(C)**, The affected side of VM; **(D)**, The affected side of BF; **(E)**, The affected side of ST; **(F)**, The contralateral side of RF; **(G)**, The contralateral side of VL; **(H)**, The contralateral side of VM; **(I)**, The contralateral side of BF; **(J)**, The contralateral side of ST; SEMG, surface electromyography; RF, rectus femoris; VM, vastus medialis; VL, vastus lateralis; BF, biceps femoris; ST, semitendinosus.

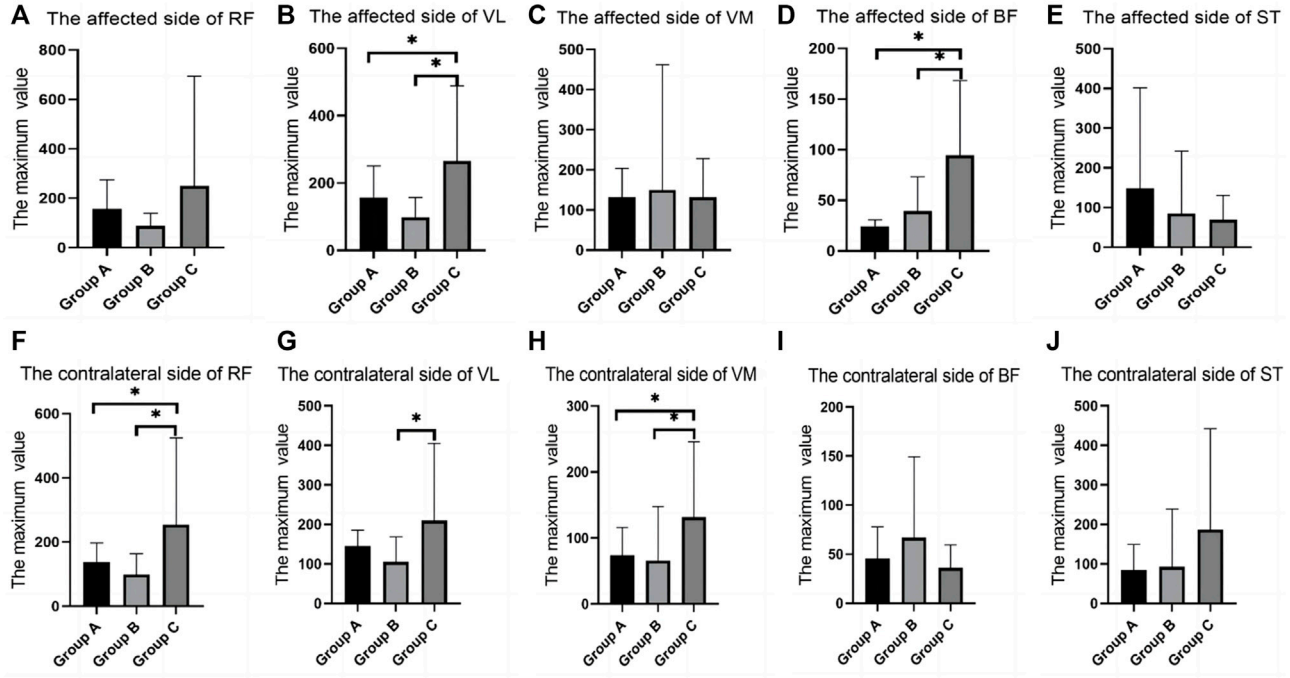

FIGURE 5
SEMG values during straight leg raising training (30°). **(A)**, The affected side of RF; **(B)**, The affected side of VL; **(C)**, The affected side of VM; **(D)**, The affected side of BF; **(E)**, The affected side of ST; **(F)**, The contralateral side of RF; **(G)**, The contralateral side of VL; **(H)**, The contralateral side of VM; **(I)**, The contralateral side of BF; **(J)**, The contralateral side of ST; SEMG, surface electromyography; RF, rectus femoris; VM, vastus medialis; VL, vastus lateralis; BF, biceps femoris; ST, semitendinosus.

TABLE 4 SEMG values during straight leg raising training (30°).

	Groups	Mean ± SD	Groups		P-value
Affected side -RF	A	157.88 ± 116.82	A	B	0.495
	B	87.37 ± 51.52	A	C	0.34
	C	249.72 ± 444.73	B	C	0.078
Contralateral side -RF	A	137.04 ± 60.52	A	B	0.546
	B	98.91 ± 65.06	A	C	**0.049**
	C	254.45 ± 269.72	B	C	**0.007**
Affected side -VL	A	157.11 ± 93.6	A	B	0.28
	B	98.22 ± 58.74	A	C	**0.036**
	C	264.84 ± 223.52	B	C	**0.001**
Contralateral side -VL	A	144.71 ± 40.56	A	B	0.39
	B	104.77 ± 63.17	A	C	0.137
	C	209.4 ± 195.43	B	C	**0.013**
Affected side -VM	A	131.68 ± 71.9	A	B	0.777
	B	149.79 ± 311.9	A	C	0.997
	C	131.88 ± 96.41	B	C	0.751
Contralateral side -VM	A	73.94 ± 41.95	A	B	0.794
	B	65.86 ± 81.77	A	C	**0.049**
	C	131.37 ± 114.17	B	C	**0.019**
Affected side -BF	A	24.36 ± 6.32	A	B	0.401
	B	39.43 ± 33.75	A	C	**0.001**
	C	94.44 ± 73.88	B	C	**0.001**
Contralateral side -BF	A	45.18 ± 32.69	A	B	0.22
	B	66.56 ± 82.5	A	C	0.575
	C	36.13 ± 23.32	B	C	0.051
Affected side -ST	A	148.64 ± 252.89	A	B	0.244
	B	84.9 ± 157.34	A	C	0.121
	C	69.38 ± 61.13	B	C	0.746
Contralateral side -ST	A	84.97 ± 64.26	A	B	0.909
	B	92.37 ± 146.48	A	C	0.096
	C	186.66 ± 255.54	B	C	0.104

SEMG, surface electromyography; RF, rectus femoris; VM, vastus medialis; VL, vastus lateralis; BF, biceps femoris; ST, semitendinosus.
That the bold values indicates the statistical significance was established.

Scottsdale, AZ, USA). The sampling frequency was 1,200 Hz. After scraping, abrading, and alcohol cleaning, the electrodes were fixed onto the patient's skin (Figure 1). SEMG signals were collected from the RF, VM, VL, BF, and ST. The signals were sampled at 1,200 Hz with a bandpass filter of 20–500 Hz. The root mean square (RMS) was used to assess muscular activity. The tasks performed during the experiment included straight leg raising (SLR) training at 30°, SLR training at 60°, ankle dorsiflexion, walking, and fast walking.

2.3 Data collection

Before the experiment began, we explained in detail what the participants needed to accomplish. Maintaining the stability of the knee while spinning is an important function of ACL (Markatos et al., 2013; Van Melick, et al., 2022); thus, we chose it as a walking task. The patient walked continuously on an elliptical curve for 5 min at the most comfortable speed. The width of the examination room was 6 meters, and the length was 8 meters. We separately

TABLE 5 The SEMG values during straight leg raising training (60°).

	Groups	Mean ± SD	Groups		P-value
Affected side -RF	A	94.67 ± 42.27	A	B	0.959
	B	91.06 ± 52.67	A	C	0.277
	C	163.19 ± 247.96	B	C	0.208
Contralateral side -RF	A	115.09 ± 62.47	A	B	0.658
	B	226.21 ± 237.17	A	C	**0.023**
	C	165.37 ± 183.06	B	C	**0.046**
Affected side -VL	A	113.1 ± 58.84	A	B	0.586
	B	165.13 ± 125.92	A	C	**0.035**
	C	135.28 ± 100.54	B	C	0.092
Contralateral side -VL	A	126.08 ± 51.23	A	B	0.566
	B	201.71 ± 131.8	A	C	**0.006**
	C	159.89 ± 107.75	B	C	**0.019**
Affected side -VM	A	144.16 ± 186.39	A	B	0.186
	B	133.62 ± 69.74	A	C	0.229
	C	126.64 ± 112.94	B	C	0.766
Contralateral side -VM	A	74.36 ± 54.74	A	B	0.733
	B	140.77 ± 154.8	A	C	0.208
	C	111.37 ± 118.88	B	C	0.074
Affected side -BF	A	32.92 ± 39.15	A	B	0.908
	B	109.87 ± 194.94	A	C	0.094
	C	70.75 ± 145.23	B	C	0.088
Contralateral side -BF	A	90.96 ± 106.28	A	B	**0.017**
	B	40.21 ± 34.47	A	C	0.695
	C	52.68 ± 66.19	B	C	**0.012**
Affected side -ST	A	65.13 ± 67.45	A	B	0.826
	B	85.7 ± 89.89	A	C	0.593
	C	76.95 ± 75.57	B	C	0.391
Contralateral side -ST	A	45.9 ± 33.91	A	B	0.579
	B	153.91 ± 225.5	A	C	0.214
	C	108.37 ± 171.34	B	C	**0.043**

SEMG, surface electromyography; RF, rectus femoris; VM, vastus medialis; VL, vastus lateralis; BF, biceps femoris; ST, semitendinosus.
That the bold values indicates the statistical significance was established.

acquire bilateral lower limbs EMG signals. And the middle 3 min EMG signal was used for analysis during the walking process. Similarly, when walking quickly, the patient was required to walk for 5 min at the fastest speed and the analysis test process was carried out for 3 min in the middle. The patient laid supine on the bed, baseline SEMG signals were measured during a state of muscle relaxation, then raised the leg 30° or 60° and dorsiflexed of the ankle, and maintained this position for 10 s (Li et al., 2020; Mitchell et al., 2022; Molina-Cárdenas et al., 2022). The experiment was repeated three times. The test is considered successful to see if the SEMG signal increases to more than 3 times the baseline signal with the active contraction movement of the muscle. The above tests are performed first on the contralateral side and then on the affected side. In the test of the maximum muscle strength of the quadriceps femoris, the patient performs a knee extension exercise in the supine position, keeping the calf hanging at the end of the bed and applying resistance at the lower end of the calf. The maximum muscle strength of the BF and ST was applied by having the patient bend the knee in the prone position and apply resistance to the crus (Dixit et al., 2022). The movement of the maximum muscle

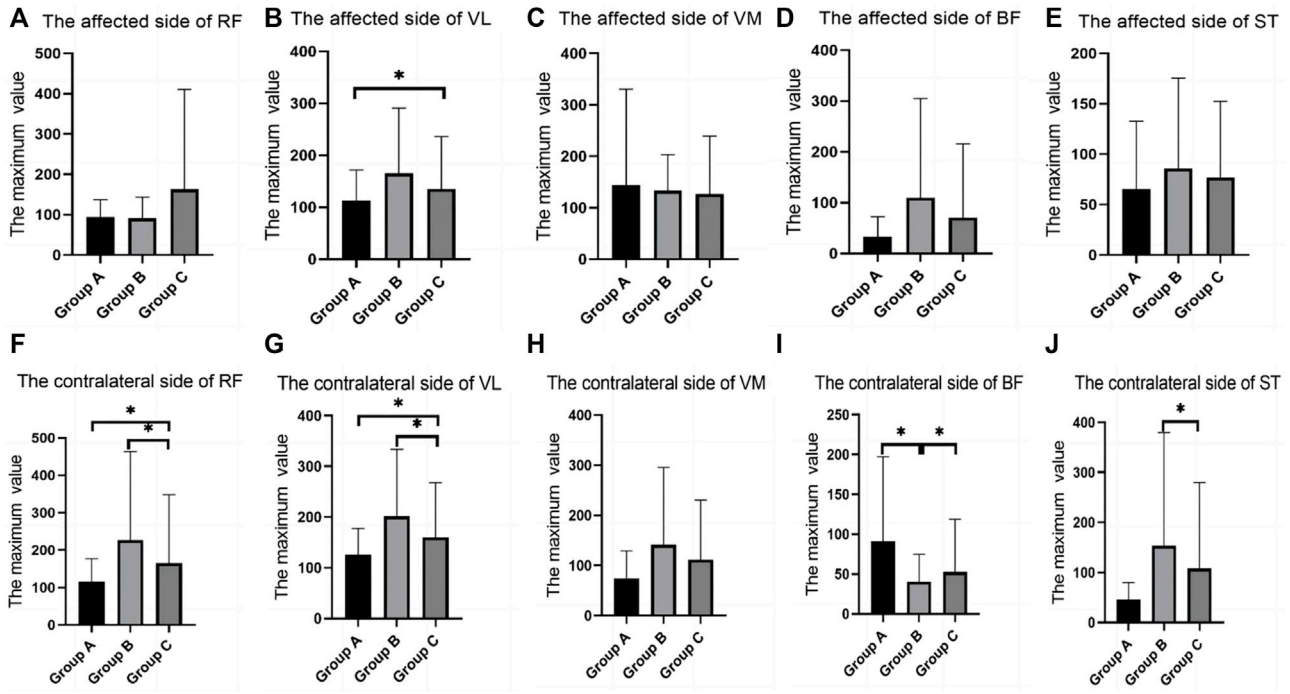

FIGURE 6
The SEMG values during straight leg raising training (60°). **(A)**, The affected side of RF; **(B)**, The affected side of VL; **(C)**, The affected side of VM; **(D)**, The affected side of BF; **(E)**, The affected side of ST; **(F)**, The contralateral side of RF; **(G)**, The contralateral side of VL; **(H)**, The contralateral side of VM; **(I)**, The contralateral side of BF; **(J)**, The contralateral side of ST; SEMG, surface electromyography; RF, rectus femoris; VM, vastus medialis; VL, vastus lateralis; BF, biceps femoris; ST, semitendinosus.

strength test is also a training exercise in the rehabilitation program. We performed an sEMG signal test on the muscle surface at maximum muscle strength to assess muscle activation during the movement, which was useful for us to evaluate the effectiveness of the activity in the rehabilitation program. The doctors and nurses accompanied the patients to ensure safety during the test.

2.4 Data process

SEMG signals were processed and analyzed using EMGworks Analysis and MATLAB R2020a (MathWorks, 3 Apple Hill Dr, Natick, MA 01760-2098). The sampling frequency was 1,200 Hz. The root mean square (RMS) value was analyzed and used to evaluate muscle activity.

2.5 Statistical methods

All data were analyzed using SPSS software (version 19.0). Data are presented as mean ± standard deviation (M±SD). The Shapiro–Wilk normality test was performed to examine the normality distribution of the data. One-way ANOVA and post-hoc Least-Significant Difference tests were used to detect statistically significant differences of the variables. Statistical significance was established at $p < 0.05$.

3 Results

In the maximum strength test, there were differences between groups in the RMS values of BF and ST on the affected side, and group A was significantly higher than group B and Group C (Table 1; Figure 2).

In the walking test, there were differences between groups in the RMS values of RF, VM, and BF. The RF of the contralateral side in group B was significantly higher than that in group C; the VM of the contralateral side in group A was significantly higher than that in group B. The BF of the affected side in group C was significantly higher than that in group A and group B (Table 2; Figure 3).

In the fast-walking test, there were differences between groups in the RMS values of RF and VL. The RF of both sides in group B were significantly higher than that in group C. The VL of the affected side in group A was significantly higher than that in group C (Table 3; Figure 4).

In the SLR training (30°) test, there were differences between groups in the RMS values of the RF, VL, VM and BF. RMS values of Group C was significantly higher than group A and Group B. VL of

TABLE 6 SEMG values during ankle dorsiflexion training.

	Groups	Mean ± SD	Groups		P-value
Affected side -RF	A	73.5 ± 74.03	A	B	0.524
	B	53.79 ± 35	A	C	0.739
	C	83.23 ± 112.88	B	C	0.249
Contralateral side-RF	A	47.17 ± 30.8	A	B	0.115
	B	80.37 ± 87.9	A	C	0.936
	C	45.61 ± 37.93	B	C	**0.046**
Affected side -VL	A	109.87 ± 107.15	A	B	0.975
	B	102.26 ± 42.8	A	C	0.403
	C	305.52 ± 1,010.29	B	C	0.32
Contralateral side -VL	A	144.67 ± 84.18	A	B	0.388
	B	110.07 ± 64.73	A	C	0.624
	C	126.2 ± 143.91	B	C	0.624
Affected side -VM	A	53.85 ± 34.8	A	B	0.141
	B	76.24 ± 47.07	A	C	0.8
	C	57.44 ± 41.33	B	C	0.133
Contralateral side -VM	A	116.9 ± 82.15	A	B	**0.016**
	B	51.34 ± 48.33	A	C	0.115
	C	77.12 ± 84.58	B	C	0.239
Affected side -BF	A	18.17 ± 22.04	A	B	0.935
	B	48.49 ± 63.91	A	C	0.394
	C	316.47 ± 1,517.49	B	C	0.381
Contralateral side -BF	A	32.34 ± 30.36	A	B	0.523
	B	37.54 ± 27.98	A	C	0.157
	C	21.41 ± 12.21	B	C	**0.019**
Affected side -ST	A	34.32 ± 27.06	A	B	0.678
	B	41.53 ± 44.89	A	C	0.909
	C	36.19 ± 57.71	B	C	0.709
Contralateral side -ST	A	27.15 ± 24.15	A	B	**0.013**
	B	69.93 ± 76.6	A	C	0.652
	C	20.01 ± 18.16	B	C	**0.001**

SEMG, surface electromyography; RF, rectus femoris; VM, vastus medialis; VL, vastus lateralis; BF, biceps femoris; ST, semitendinosus.
That the bold values indicates the statistical significance was established.

the contralateral side of VL in group B was significantly decreased when compared with that in group C (Table 4; Figure 5).

In the SLR training (60°) test, there were differences in RMS values for RF, VL, BF, and ST between groups. The contralateral RF in group C was shown to be significantly larger than that in group A and smaller than that in group B. The contralateral VL in group C was significantly larger than that in group A and smaller than that in group B. The VL of the affected side in group A was significantly decreased when compared with that in group C. The ST of contralateral side in group B was significantly larger than that in group C. The VL of the affected side in group A was significantly decreased when compared with that in group C. The contralateral BF in group B was significantly decreased when compared with that in group A and group C (Table 5; Figure 6).

In the ankle dorsiflexion training test, there were differences between groups in the RMS values of RF, BF, and ST on the contralateral side, and group B was significantly higher than group C. The RMS value of contralateral side of VM in group A was significantly larger than that in group C (Table 6; Figure 7).

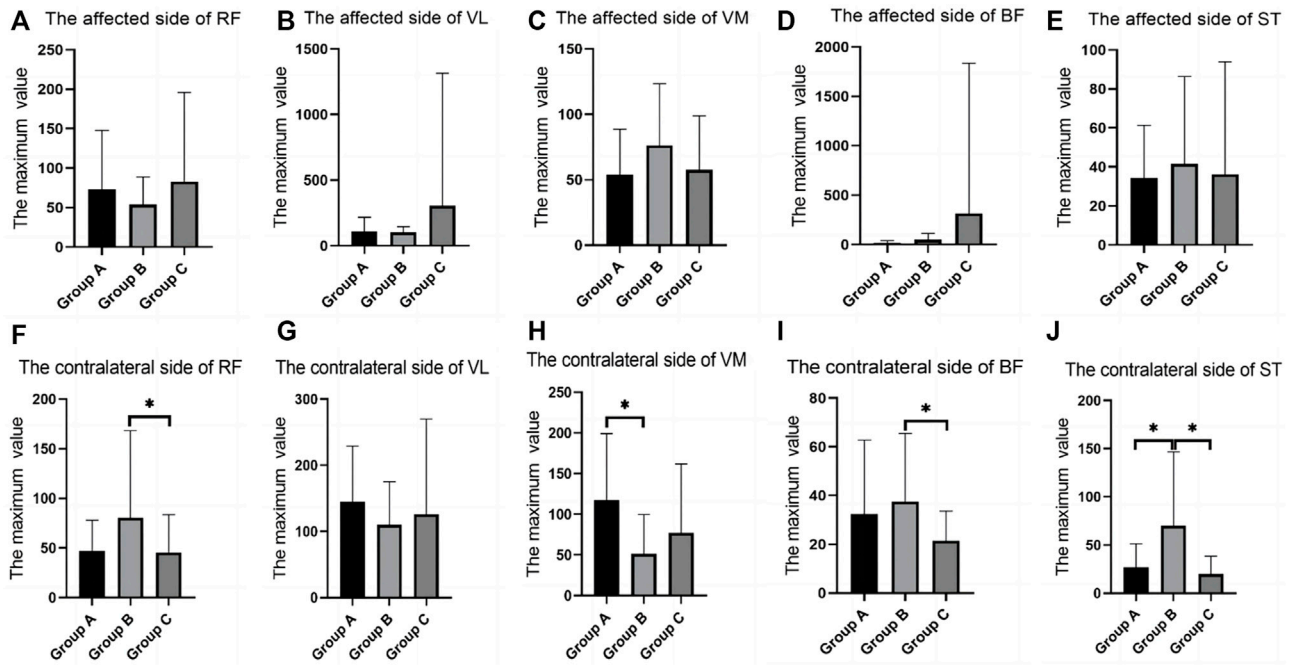

FIGURE 7
SEMG values during ankle dorsiflflexion training. **(A)**, The affected side of RF; **(B)**, The affected side of VL; **(C)**, The affected side of VM; **(D)**, The affected side of BF; **(E)**, The affected side of ST; **(F)**, The contralateral side of RF; **(G)**, The contralateral side of VL; **(H)**, The contralateral side of VM; **(I)**, The contralateral side of BF; **(J)**, The contralateral side of ST; SEMG, surface electromyography; RF, rectus femoris; VM, vastus medialis; VL, vastus lateralis; BF, biceps femoris; ST, semitendinosus.

4 Discussion

This study explored the muscles' SEMG characteristics of patients with ACL injuries in different rehabilitation phases. The results showed that the maximum muscle strength has a significant tendency to decrease, and the attenuation of each muscle during different movements was also different. However, muscle strength recovery in BF and ST are significantly poorer. Although muscle strength and symmetry of the bilateral lower extremities play an important role in patient recovery, muscle atrophy is common in patients who require local fixation. Factors related to muscle size and strength loss after ACLR remain unclear. The decrease in the maximum muscle strength of the BF and ST may be due to the influence of the operation. Furthermore, some studies have been conducted on age, preoperative and postoperative activity levels, the time elapsed since surgery, ACL grafts, rehabilitation programs, concurrent operative procedures, and other methods have been used to define muscle size and function. Muscle atrophy appears to occur regardless of the type of graft used to reconstruct the ACL and often occurs bilaterally (Papannagari et al., 2006). The pattern of muscle strength decline is not the same for different muscles, Li et al. showed that Among the VL, VM, and RF, the recovery rate of VL is the slowest (Li et al., 2020). Kim et al. also found significant differences in the recovery rates of extensor muscles and flexor muscles of knee (Kim et al., 2015).

Changes in the biomechanics of the lower extremities are also important factors that lead to muscle changes. Surgery cannot completely achieve the anatomical reduction of the ACL, which can lead to a change in torque. The symmetry of both legs changes due to the change in torque, especially during contralateral movements. Numerous studies have evaluated the effect of common musculoskeletal impairments after ACLR on lower-extremity movement asymmetries. Some studies have suggested that patients with high quadriceps strength symmetry demonstrated lower limb movement symmetry and greater functional performance than those with low quadriceps strength symmetry (Lewek et al., 2002; Palmieri-Smith and Lepley, 2015). Moreover, many studies have found that the biomechanical characteristics of both lower extremities are still inconsistent after ACLR. Roewer et al. (2011) reported asymmetric knee angles and moments after ACLR in participants with symmetrical quadriceps strength.

Additionally, Curran et al. (2018) reported that although quadriceps strength improved after reconstruction, there were still asymmetric biomechanical parameters of the knee. Shi et al. (2019) reported that the isometric quadriceps strength of the injured leg was significantly lower than that of the uninjured leg. Knee flexion angles and knee extension moments were smaller in the injured leg than in the uninjured leg during the loading response and mid-stance phases. Asymmetry in the maximum knee flexion angle during the loading response and mid-stance phases was significantly correlated with asymmetry in the isometric quadriceps strength. Isometric quadriceps strength was also significantly correlated with the asymmetry in the peak knee extension moment during the midstance phase. The authors believed that individuals receiving ACLR demonstrate asymmetry in knee movement in the sagittal

plane. Isometric quadriceps strength asymmetry was significantly correlated with the asymmetry in knee flexion angles during the early stance phase and with the knee extension moments during the midstance phase. Palmieri-Smith and Lepley (2015) also suggested that decreased QF strength is associated with altered trunk biomechanics of the lower extremity and sagittal plane during landing tasks after ACLR. However, the relationship between QF muscle strength and lateral trunk control after ACLR during landing has not been examined. Furthermore, weakness in the hip abductor (HA) muscles is often associated with a dynamic valgus of the knee during landing activities, a pattern of motion associated with both primary and secondary risks of ACL injury (Petersen et al., 2014). Therefore, rehabilitation programs should emphasize eccentric exercise to improve neuromuscular control.

This study showed that different muscles exhibited an obvious decrease in muscle activation under different exercise conditions, which closely related to the patient's sports ability. For example, the muscle strength of the contralateral side of the RF, the contralateral side of the VM, and the affected side of the BF changed during normal walking and that of the affected side of the RF, the contralateral side of the RF, and the affected side of the VL changed during a fast walking movement. ACL structures also contain mechanoreceptors, injuries of which cause afferent block-mediated dysregulation of motor control of the spine and spinal cord, directly affecting neuromuscular control of the knee. ACL injury can cause changes in biomechanical function, proprioceptive function, and strategy of motor control (Van Melick et al., 2016). Begalle et al. (2012) quantified the activity level of VM, VL, MH, and BF muscles by SEMG. In exercises such as the single-limb squat, the transverse-lunge, lateral-lunge, etc, the extensor and flexor muscles of the knee show a marked difference. Therefore, the muscle training method must refer to the target motion modes and evaluate the quality of movement. The focus on quality evaluation of movement is not only a risk factor for a second ACL injury, but also an important factor for motor rehabilitation.

5 Limitation

Only 60 patients participated in this study, and the limited sample size reduced the informative value of the study. The lack of follow-up also reduces the representativeness of the results. Therefore, we will continue to expand the sample size of this study, achieve long-term follow-up of patients, and provide more evidence to validate the results.

6 Conclusion

Overall, the muscle activation of various muscles differ in several movements. Therefore, according to the test results under different exercise conditions, we may find which muscle requires more training and improves the functional status of patients after surgery. We suggest developing a rehabilitation program based on target orientation. At different phases of rehabilitation, testing the activation of muscles under different training tasks, and the training tasks and the intensity of the training tasks are selected based on this. We speculate that there will be better rehabilitation.

Data availability statement

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

Ethics statement

The studies involving human participants were reviewed and approved by the Medical Ethics Committee of Binzhou Medical University Hospital approved this study (2021-S019-01). The patients/participants provided their written informed consent to participate in this study.

Author contributions

WL and SQ provided the conception/design, and SW, HZ, ZC, XW, and YZ performed the experiments and assembled the data. HC and ZC conducted the data analysis and interpretation. HC and ZC wrote the manuscript, and WL and SQ revised it. HC, ZC, SW, HZ, ZC, XW, YZ, SQ, and WL finally approved the manuscript. All authors read and approved the final version of the manuscript.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Effect of kinesio taping intervention on the muscle strength and balance of college basketball players with functional ankle instability

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Objective: The aim of this study was to investigate the effects of acute Kinesio Taping (KT) intervention on the muscle strength and balance ability of college basketball players with functional ankle instability (FAI).

Methods: Thirty college basketball players with FAI were treated with acute KT to test the changes in their muscle strength and balance ability.

Results: After acute KT intervention, the ankle dorsiflexion moment and the ankle plantar flexion moment increased by 34% and 19.9%, respectively. The stable plane test with the subjects' eyes open decreased by 1%, whereas that with the subjects' eyes closed decreased by 1.1%. The swaying environment test with the subjects' eyes open increased by 2.4%. The swaying plane test with the subjects' eyes open increased by 5.1%, whereas that with the subjects' eyes closed increased by 16.2%. The swaying environment test with the subjects' eyes open plus the use of a plane increased by 12.1%.

Conclusion: KT can increase the isokinetic strength of the ankle dorsum muscle and plantar flexion of college basketball players with FAI. The effect of KT in the static balance test was weaker than that in the dynamic balance test. The findings indicate that KT can significantly improve the balance ability of college basketball players with FAI during dynamic sports.

KEYWORDS

functional ankle instability, college basketball player, kinesio taping, isokinetic muscle strength, balance

1 Introduction

An ankle sprain is the main ankle injury in sports. The incidence of ankle injury is 76.7%, accounting for 10%–17% of all joint injuries in college athletes (Kim et al., 2022; Serra-Añó et al., 2021; D'Hooghe et al., 2020). Lateral ankle joint injury is one of the most common injuries (Doherty et al., 2017; Herzog et al., 2019; Alshahrani and Reddy, 2022). Studies have shown that functional ankle instability (FAI) is common in people who play jumping and turning sports, such as basketball, volleyball or soccer (Lin et al., 2021a). Approximately 70%–80% of basketball players develop FAI after their first ankle sprain (Santos and Liu, 2008). FAI leads to decreased postural control ability, which manifests as poor muscle

strength (Mao et al., 2021), balance (Doherty et al., 2014), proprioception (Alghadir et al., 2020; Xu et al., 2022) and neuromuscular control (Huang et al., 2021). Falling may also increase the risk of a second sprain (Simpson et al., 2019).

Kinesio Taping (KT) was invented by Dr. Kenso Kase in Japan (Williams et al., 2012). The application of KT supports and relaxes muscles, ligaments and fascia, as well as other soft tissues. Tai Chi combined with KT significantly improves the dynamic and static balance ability of FAI football players (Li et al., 2022a). Furthermore, KT increases dynamic postural control in FAI patients after ankle muscle fatigue (Li et al., 2022b). KT application also improves the lower limb stability of patients with chronic ankle instability (CAI) (Yu et al., 2021). However, some studies suggest that KT has no positive effect on postural control and functional activities (Nunes et al., 2021).

Injuries to the ankles are usually caused by inversion (Woods et al., 2002; Doherty et al., 2014). Hence, increasing the strength around the ankle joint can improve the dynamic postural stability of the body during landing movement (Dewar et al., 2019). Increasing ankle muscle strength *via* KT intervention may reduce the recurrence of sprain in basketball (Shilun, 2011). KT application to college basketball players with FAI has recently been the focus of muscle strength and balance ability research, and the existing research on KT on muscle strength and balance ability is still controversial. The results of some studies indicate that KT does not seem to increase muscle strength or improve the balance ability of subjects (Shields et al., 2013; Bailey and Firth, 2017; Martonick et al., 2020). However, other studies have shown that KT plays a certain role in improving the muscle strength and proprioception of injured people (Williams S et al., 2012; Lin et al., 2021b; Mao et al., 2021; Biz et al., 2022). In this study, the aforementioned research scope is clarified by analysing the changes in muscle strength and balance ability through KT intervention and then evaluating its effect on the ankle of college student basketball players with FAI. The KT intervention was also conducted to improve muscle strength and balance ability and consequently derive the application's theoretical basis. The research hypothesis is stated as follows: KT intervention can improve the ankle isokinetic muscle strength and balance ability of college basketball players with FAI.

2 Materials and methods

2.1 Research object

This study was approved by the Human Testing Ethics Committee of Chengdu Sport University [2022] No. 3, and 30 college basketball players who met the standards were selected for this research. The age range of the subjects was 21.5 ± 1.2 years, their height range was 179.33 ± 7.56 cm, and their weight range was 78.9 ± 9.76 kg. Regarding the Cumberland CAI tool (CAIT), the left side was 24.07 ± 6.27, and the right side was. 22.4 ± 5.59), with 13 and 17 patients presenting left ankle points (≤24) and right ankle points (≤24), respectively.

2.2 Selection of samples

The diagnostic criteria were as follows (Santos and Liu, 2008): 1) two or more unilateral varus sprains of the ankle joint; 2) subjective ankle instability that occurred twice or more in the past 6 months; and 3) negative anterior drawer test of the talus.

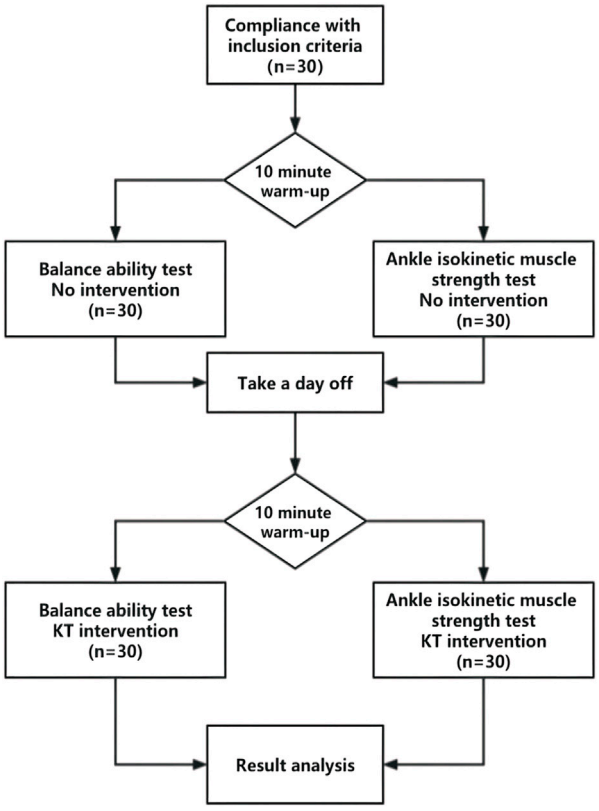

FIGURE 1
Experimental design flow chart.

Included in the standard were the following items (Delahunt et al., 2010; Rein et al., 2020): 1) repeated ankle sprain more than twice; 2) aged 18 to 25; 3) most recent sprain was 30 days ago; 4) completed the ankle instability questionnaire by using CAIT (Wright et al., 2017), with a score of ≤24 points for the unilateral ankle joint; 5) a negative result in the anterior ankle drawer test; 6) no history of fracture or surgery of the lower limbs; and 7) informed that the study was based on the Declaration of Helsinki, with informed consent forms signed.

The exclusion criteria were as follows (Kunugi et al., 2018): 1) evident dislocation and fracture; 2) simultaneous severe sprain of both ankle joints; 3) other neurological diseases that may affect balance and muscle strength; and 4) skin allergy to KT.

2.3 Research methods

2.3.1 Experimental method

Subjects were tested before and after the KT intervention. In the first experiment (i.e., before the KT intervention), the balance ability and lower limb muscle strength of all subjects were tested. In the second experiment (i.e., after KT intervention), the balance ability and lower limb muscle strength of all subjects were tested. The interval between the two experiments was 1 day, during which the

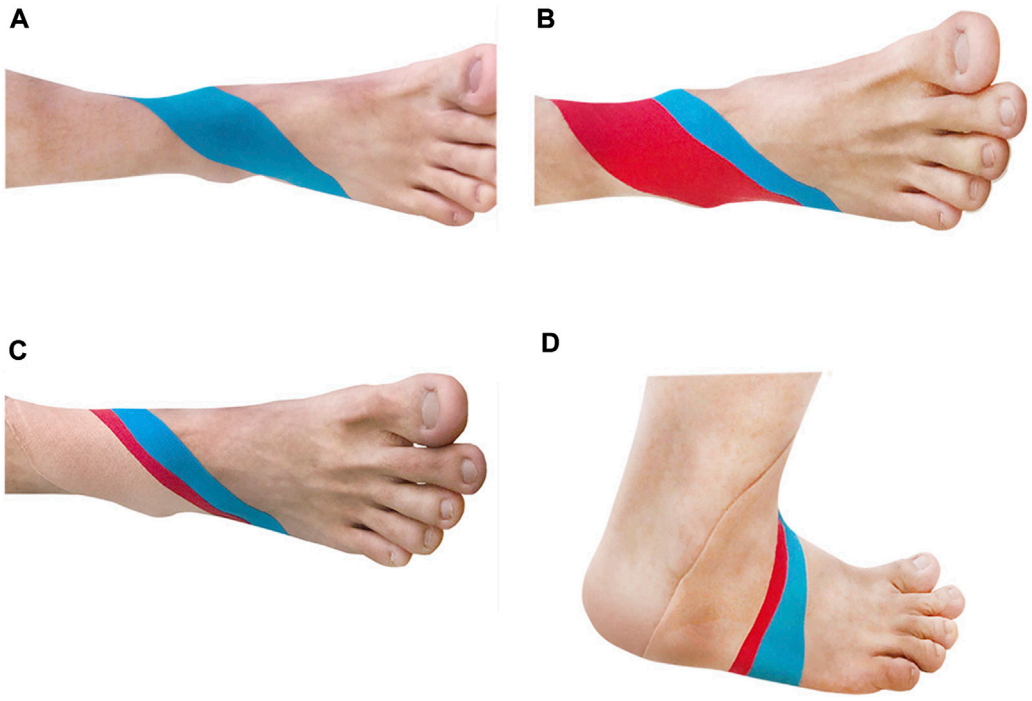

FIGURE 2
KT group **(A)** The first KT; **(B)** The second kT **(C)** The third KT; **(D)** Schematic diagram after completion.

subjects were required to maintain normal sleep and were not allowed to perform any muscle strength or balance training of the lower limbs (Figure 1).

2.3.2 Stick firm method

According to John et al. (2018), for the first KT intervention, the anchor point should be set below the medial malleolus to prevent ankle sprain. In this method, an experimenter stretches the KT forward and pastes it along the position of the talus whilst simultaneously pressing the subject's thigh against the ball of the foot (i.e., the subject's foot is in a dorsiflexion position). Moderate stretching tension (approximately 30%) was applied to the KT through the lateral malleolus and terminated at the sole (Figure 2A). For the second kT intervention, the experimenter reversed pastes the KT along the first path with maximum tension (approximately 50%). The KT, passing from the outside of the foot, is stretched along the front of the ankle, and the pasting is terminated at the inside of the tibia (i.e., the KT should be one-third shorter than the original measurement length) (Figure 2B). If necessary, a third KT intervention may be adopted. The anchor point is set at the bottom of the foot. Without any tension to the KT, the experimenter, starting from the middle position, passing from the outside and front of the foot and bypassing the ankle joint, places the tail-end of the KT on the inside of the tibia. Then, the other end of the KT is placed by the experimenter on the inside of the foot, straining through the medial malleolus, then pulling back and terminating at the end of the first 2 KTs. The third KT intervention is usually conducted to allow the first and second kT interventions to fit well into the skin (Figure 2C). Figure 2D presents an image of a completed KT intervention.

2.3.3 Balance ability test

The Smart Equi Test balance training tester developed in the United States was used, and the sensory integration experiment (SOT) was selected to perform six tests, which can be described as follows: 1) subjects with eyes open and floor and visual environment stable; 2) subjects with eyes closed and floor stable; 3) subjects with eyes open, floor stable and visual environment mobile; 4) subjects with eyes open, floor mobile and visual environment stable; 5) subjects with eyes closed and floor mobile; and 6) subjects with eyes open and floor and visual environment mobile. The tests were scored between 100 points (stable) and 0 points (fall). The balance ability index was taken as the test index; the higher the score is, the stronger the balance ability (Pickerill and Harter, 2011). Table 1 presents a detailed overview of the SOT (Delahunt et al., 2010).

2.3.4 Ankle isokinetic muscle strength test

Flexion and extension tests (45°/s, 15 times) were performed on the affected ankle joints of 30 subjects by using the Swiss CONTRE isokinetic tester. The subjects were placed in the supine position during the test. Before the experiment, the subjects were allowed to warm up for 10 min. Then, the experimenter verbally stimulated the subjects to ensure that they could complete the whole experimental process as scheduled.

Peak torque was used as the test index. The maximum torque value of the joint muscle group, calculated during the ankle isokinetic test, was taken as the muscle strength level of the subject (Cheng et al., 2019; Cheng and Jiang, 2020).

TABLE 1 Overview of the six SOT tests.

Test project	Visual environment		System response	
	Eyes	Floor	Prejudice	Apply
1	Open	stable	—	proprioception
2	Closed	stable	visual sense	proprioception
3	Open	stable	visual sense	proprioception
4	Open	instability	proprioception	visual sense and vestibular
5	Closed	instability	Proprioception and visual sense	vestibular
6	Open	instability	Proprioception and visual sense	vestibular

Note: Test 1: Eyes Open, Floor and Visual Environment Stable; Test 2: Eyes Closed, Floor Stable; Test 3: Eyes Open, Floor Stable, Visual Environment Mobile; Test 4: Eyes Open, Floor Mobile, Visual Environment Stable; Test 5: Eyes Closed, Floor Mobile; Test 6: Eyes Open, Floor and Visual Environment Mobile.

TABLE 2 Changes in isokinetic muscle strength of the affected ankle joint before and after KT intervention.

MG	NTPM (N·m)	KTPM (N·m)	PC (%)	t	*p*	95% CI
DF	45.0 ± 22.3	60.3 ± 21.9	34.0	−2.197	0.045*	−30.142 ~ −0.364
PF	48.8 ± 23.9	58.5 ± 22.3	19.9	−2.223	0.039*	−18.814 ~ −0.566

Note: MG, muscle group; NTPM, no taping intervention peak moment; KTPM, kinesio taping intervention peak moment; PC, percentage change; Nm, Newton m (moment unit); PF, plantarflexion; DF, dorsiflexion; Values are means ± standard deviation (SD); Significant differences ($p < 0.05$).

TABLE 3 Changes in balance ability before and after KT intervention.

Test project	PBI (score)	ABI (score)	PC (%)	*t*	*p*	95% CI
1	95.5 ± 1.2	94.5 ± 2.3	−1.0	1.871	0.082	−0.148–2.174
2	93.3 ± 2.6	92.2 ± 3.0	−1.1	1.321	0.208	−0.744–3.130
3	92.3 ± 2.5	94.6 ± 2.3	2.4	−3.883	0.002*	−3.684 ~ −1.062
4	85.6 ± 7.3	90.0 ± 4.9	5.1	−5.405	0.001*	−6.211 ~ −2.682
5	62.1 ± 17.9	72.2 ± 10.7	16.2	−3.309	0.005*	−16.70 ~ −3.567
6	73.1 ± 15.0	82.0 ± 5.5	12.1	−3.242	0.006*	−14.76 ~ −3.008

Note: PBI, Pre-intervention balance index; ABI: After-intervention balance index; PC, percentage change; Test 1: Eyes Open, Floor and Visual Environment Stable; Test 2: Eyes Closed, Floor Stable; Test 3: Eyes Open, Floor Stable, Visual Environment Mobile; Test 4: Eyes Open, Floor Mobile, Visual Environment Stable; Test 5: Eyes Closed, Floor Mobile; Test 6: Eyes Open, Floor and Visual Environment Mobile.

2.4 Data statistics and analysis

The measurement data were processed as the mean ± standard deviation by using SPSS 20.0 statistical software. For measures conforming to normal distributions, the data of two measurements were tested by paired sample *t*-test, with the significance level set to $\alpha = 0.05$.

3 Results

3.1 Isokinetic muscle strength of the ankle joint

After KT intervention, the peak moment of ankle dorsiflexion increased by 34% from 45.0 ± 22.3 to 60.3 ± 21.9 ($p = 0.045$). The moment of ankle plantar flexion increased by 19.9% from 48.8 ± 23.9 to 58.5 ± 22.3 ($p = 0.039$). The test results suggest the ability of the KT intervention to improve the ankle plantar flexion and dorsi flexion muscle strength of college basketball players with FAI. The difference was statistically significant (Table 2).

3.2 Balance of the ankle joint

According to the SOT test results, the equilibrium index of condition 1 decreased by 1% from 95.5 ± 1.2 to 94.5 ± 2.3 ($p = 0.082$), and the difference was statistically insignificant (Table 3). The balance index of condition 2 decreased by 1.1% from 93.3 ± 2.6 to 92.2 ± 3.0 ($p = 0.208$), and the difference was also statistically insignificant. The balance index of condition 3 increased by 2.4% from 92.3 ± 2.5 to 94.6 ± 2.3 ($p = 0.002$),

with a statistically significant difference. These results indicate the ability of KT to improve the vestibular sensation of the ankle joint. In addition, the balance index of condition 4 increased from 85.6 ± 7.3 to 90.0 ± 4.9, which was 5.1% higher (p = 0.001), and the difference was statistically significant. Therefore, KT can improve the proprioception of the affected ankle joint. The balance index of condition 5 increased by 16.2% from 62.1 ± 17.9 to 72.2 ± 10.7 (p = 0.005), with a statistically significant difference. Hence, KT can significantly improve the proprioception of the affected ankle joint under visual deprivation conditions. The balance index of condition 6 increased by 12.1% from 73.1 ± 15.0 to 82.0 ± 5.5 (p = 0.006), and the difference was statistically significant. Thus, KT can improve balance ability even when both vestibular and proprioception are disturbed. In contrast, in conditions 1 and 2, the difference was statistically insignificant after intervention, as vision was not disturbed or only vision was shielded.

4 Discussion

The purpose of this study was to verify whether KT intervention can help improve the ankle isokinetic muscle strength and balance ability of college basketball players with FAI. After KT intervention, the performance of the subjects who participated in the ankle joint isokinetic muscle strength test significantly improved. In the static balance test, the difference in the results of the KT intervention was statistically insignificant. However, in the dynamic balance test, KT intervention significantly improved the balance ability of college basketball players with FAI.

4.1 Effect of KT on the ankle isokinetic muscle strength of college basketball players with FAI

This study verified the hypothesis that KT intervention can improve ankle isokinetic muscle strength in college basketball players with FAI. The obtained results showed a 34% and 19.9% increase in ankle extensor and flexor peak moments, respectively, which was similar to previous findings (Biz et al., 2022). Interestingly, although Muñoz-Barrenechea et al. (2019), reported that KT effectively improved the muscle strength of subjects with ankle instability, a few other researchers disagree. Nunes and de Noronha, 2019 applied KT to the calf triceps of 30 athletes and found no difference before and after the intervention. Ruoni (2021) found that KT could not significantly improve ankle dorsiflexion range of motion, lower limb muscle strength or subjective sensation in CAI patients within a short time. The conflicting results may be related to the different sticking methods and testing methods used by the researchers. In this study, the ability of KT to improve the isokinetic muscle strength of the subjects' ankle joint may be explained by the elastic retraction force of the KT itself, producing traction on the subcutaneous tissue, thus causing neuromuscular stimulation and increasing muscle strength. On the basis of this analysed mechanism, KT may have enhanced the contraction ability of the damaged muscle, reduced the pain caused by muscle overextension, relieved the probability of spasm caused by muscle fatigue, improved the stability of the injured joint and surrounding soft tissue, prevented the abnormal work of the muscle from causing movement disorders of the normal joint and maintained the normal range of motion of the joint. These mechanisms can jointly reduce pain perception and increase muscle strength (Sarvestan and Svoboda, 2019). In addition, some researchers have reported the immediate effect of KT on muscle strength. For example, the tension of the pasting cloth can mechanically pull the muscle fascia, stimulate the contraction of weaker muscles and indirectly increase muscle strength (Merino-Marban et al., 2021; Li et al., 2022a).

4.2 Effects of KT on the balance ability of college basketball players with FAI

This study verified the research hypothesis that KT intervention can improve the balance ability of college basketball players with FAI. The equilibrium indices of conditions 3, 4, 5, and 6 increased by 2.4%, 5.1%, 16.2% and 12.1%, respectively, which is consistent with the findings of previous research (Buchanan et al., 2008; Bicici et al., 2012). Ruoni (2021) also found that KT can improve ankle proprioception and lower limb dynamic balance ability in CAI patients in a short period. However, past findings were generally inconsistent. Yazici et al. (2015) found that KT intervention in the ankle joint of stroke patients could not change their static balance. Esposito et al. (2021) found that KT could not improve the static and dynamic balance of healthy semiprofessional football players. The varying results may be caused by different experimental subjects or intervention methods. In this study, the KT intervention was able to improve balance ability, which may be explained by the following: after KT is applied to the skin surface, sensory input information of the skin is generated, and proprioceptors may appear as collateral injury after ankle injury. Furthermore, the use of KT can stimulate skin effectors to enhance sensory afferent information; consequently, additional sensory and perceptual pathways are formed in the central nervous system, and the path of the reflex arc is increased, which improves the central nervous control of the periphery whilst enhancing stability (Cheng et al., 2017; Wang et al., 2018). Jackson et al. (2016) also proposed the application of KT to increase proprioceptive function and improve balance performance. The results of this research indicate that balance scores can be significantly improved compared with before intervention, especially when visual and vestibular perception is unfavourable, confirming the conclusions of Ruoni (2021) and Jackson et al. (2016).

4.3 Limitations

This study, which entailed a small sample with a no-placebo control, encountered limitations. Future researchers may use more subjects to compare differences in age, sex and other factors. In addition, different taping methods have not yet been applied to patients to determine whether different taping methods have varying effects on college basketball players with FAI. Future studies may also consider adding other measures (i.e., neuromyography) and implementing longer follow-up investigations. Non-etheless, the mechanism related to KT's improvement of muscle strength and the balance ability of FAI patients must be explored to generate more indicators.

5 Conclusion

KT can increase the isokinetic strength of the ankle dorsalis and plantaris flexion in college basketball players with FAI. The KT intervention presented a negligible effect on lower limb stability in patients with FAI during static balance tests, but it significantly improved the balance ability of FAI patients in the dynamic balance test.

Data availability statement

The original contributions presented in the study are included in the article/supplementary material, further inquiries can be directed to the corresponding authors.

Ethics statement

The studies involving human participants were reviewed and approved by This study requested informed consent and was approved by the Ethics Committee of Chengdu Sport University [2022] No. 3. The patients/participants provided their written informed consent to participate in this study.

Author contributions

RL, RQ, YT, KW, HL, and LC contributed to conception and design of the study. RL and RQ organized the database. YT performed the statistical analysis. RL wrote the first draft of the manuscript. RQ, YT, KW, LC, and HL wrote sections of the manuscript. All authors contributed to manuscript revision, read, and approved the submitted version.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Evaluating the risk of knee osteoarthritis following unilateral ACL reconstruction based on an EMG-assisted method

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Objective: Anterior cruciate ligament reconstruction (ACLR) cannot decrease the risk of knee osteoarthritis after anterior cruciate ligament rupture, and tibial contact force is associated with the development of knee osteoarthritis. The purpose of this study was to compare the difference in bilateral tibial contact force for patients with unilateral ACLR during walking and jogging based on an EMG-assisted method in order to evaluate the risk of knee osteoarthritis following unilateral ACLR.

Methods: Seven unilateral ACLR patients participated in experiments. The 14-camera motion capture system, 3-Dimension force plate, and wireless EMG test system were used to collect the participants' kinematics, kinetics, and EMG data during walking and jogging. A personalized neuromusculoskeletal model was established by combining scaling and calibration optimization. The inverse kinematics and inverse dynamics algorithms were used to calculate the joint angle and joint net moment. The EMG-assisted model was used to calculate the muscle force. On this basis, the contact force of the knee joint was analyzed, and the tibial contact force was obtained. The paired sample *t*-test was used to analyze the difference between the participants' healthy and surgical sides of the participants.

Results: During jogging, the peak tibial compression force on the healthy side was higher than on the surgical side ($p = 0.039$). At the peak moment of tibial compression force, the muscle force of the rectus femoris ($p = 0.035$) and vastus medialis ($p = 0.036$) on the healthy side was significantly higher than that on the surgical side; the knee flexion ($p = 0.042$) and ankle dorsiflexion ($p = 0.046$) angle on the healthy side was higher than that on the surgical side. There was no significant difference in the first ($p = 0.122$) and second ($p = 0.445$) peak tibial compression forces during walking between the healthy and surgical sides.

Conclusion: Patients with unilateral ACLR showed smaller tibial compression force on the surgical side than on the healthy side during jogging. The main reason for this may be the insufficient exertion of the rectus femoris and vastus medialis.

KEYWORDS

anterior cruciate ligament reconstruction, knee osteoarthritis, tibial contact force, EMG-assisted model, gait

1 Introduction

Anterior cruciate ligament (ACL) injury is one of the most common sports-related injuries (Agel et al., 2005; Yu and Garrett, 2007), and anterior cruciate ligament reconstruction (ACLR) cannot decrease the risk of knee osteoarthritis. A study suggested that patients with a single ACL rupture have a 10%–13% risk of developing knee osteoarthritis 10 years after injury (Oiestad et al., 2009). This risk increases from 21% to 48% in patients with a torn meniscus associated with ACL rupture (Oiestad et al., 2009). ACLR surgery is a common treatment for ACL rupture, but previous studies have shown that ACLR surgery cannot reduce the incidence of knee osteoarthritis (Lohmander et al., 2007; Barenius et al., 2014; Cinque et al., 2018; Grassi et al., 2022). A study showed that the incidence of knee osteoarthritis on the ACLR surgical side increased by three times compared with the contralateral side (Barenius et al., 2014). Studies have also shown that, on average, 50% of patients with ACL or meniscus tears will develop knee osteoarthritis after 10–20 years (Lohmander et al., 2007). Despite the high incidence of knee osteoarthritis after ACLR, the exact cause remains unclear.

There will be a series of changes in biomechanical characteristics for ACLR patients in daily activities, which may increase the risk of knee osteoarthritis. Studies conducted on patients who have undergone ACLR have reported a decrease in peak knee flexion angles (Roewer et al., 2011; Hall et al., 2012; Webster et al., 2012; Saxby et al., 2016), knee flexion moments (Bush-Joseph et al., 2001; Lewek et al., 2002; Webster et al., 2005; Di Stasi et al., 2013; Zabala et al., 2013), and external knee adduction moments (Webster and Feller, 2012; Zabala et al., 2013; Kaur et al., 2016) during prolonged walking or jogging after the surgery. Additionally, ACLR patients' muscle activation strategy also changed at the same time, which is known in rehabilitation field as arthrogenic muscle inhibition. After ACLR, quadriceps activation failure caused by neural inhibition occurs (Sonnery-Cottet et al., 2019). Mechanisms for this phenomenon include changes in muscle resting motor thresholds, in the discharge of articular sensory receptors (Rice and McNair, 2010) and in the cortical activity (Baumeister et al., 2008; Baumeister et al., 2011). Arthrogenic muscle inhibition also shows in the daily movement. Several studies' results suggested that there is an increase in hamstring activation (Knoll et al., 2004; Mantashloo et al., 2020) and a decrease in quadriceps femoris activation (Leporace et al., 2016) among ACLR patients during walking. The altered biomechanics in movement will significantly affect the mechanical environment of the knee joint, which may increase the risk of knee osteoarthritis. Bone loads are crucial to the growth and remodeling process of joint tissues (Beaupre et al., 2000; Carter et al., 2004). There is a dynamic balance between the reabsorption of bone matrix by osteoclasts and the formation of new bone by osteoblasts. However, this dynamic balance can be disrupted by abnormal loads, which may cause degenerative changes in joint tissue (Radin and Paul, 1971). Therefore, assessing tibial contact forces can improve understanding of knee osteoarthritis mechanisms.

The contribution of muscle forces cannot be ignored when assessing the tibial contact force, but evaluating muscle forces non-invasively during movement has always been a challenge because the human musculoskeletal system has redundancy (Crowninshield and Brand, 1981; Lloyd and Besier, 2003). Two studies estimated tibial contact forces for ACLR patients, and both used the EMG-driven method to evaluate muscle forces during a gait task (Saxby et al., 2016; Wellsandt et al., 2016). Wellsandt et al. (Wellsandt et al., 2016) suggested that patients who get knee osteoarthritis 5 years after ACLR surgery have a smaller tibial compression force during walking. Saxby et al. (Saxby et al., 2016) obtained the consistent result that the patients in the ACLR group show a smaller tibial compression force than the healthy group during walking, jogging, and cutting. Although the EMG-driven method uses EMG signal as input to the musculoskeletal simulation to solve the muscle redundancy problem, it still has some limitations like the difficult measurement of EMG signal of deep muscles and the large error with experimental results caused by signal noise (Sartori et al., 2014; Hoang et al., 2019). The EMG-assisted method combines the optimization and EMG-driven methods, which can significantly improve the experimental joint moment tracking accuracy than current EMG-driven models (Sartori et al., 2014; Pizzolato et al., 2015; Hoang et al., 2019).

The purpose of this study was to compare the difference in bilateral tibial contact force for patients with unilateral ACLR during walking and jogging based on an EMG-assisted method to evaluate the risk of knee osteoarthritis. We hypothesized that: 1) ACLR patients' tibial compression force on the surgical side would be smaller than on the healthy side during walking and jogging; 2) there would be difference in muscle forces around the knee joint between the surgical and healthy sides during walking and jogging; 3) there would be difference in lower limb kinematics between the surgical and healthy sides during walking and jogging.

2 Materials and methods

2.1 Participants

Seven ACLR participants (three males and four females, height: 1.72 ± 0.13 m, weight: 67.4 ± 14.3 kg, 1.5–3.5 years post-ACLR, the surgical side of the five participants was the dominant side) participated in this study. None of the participants had other lower limb injuries and did not engage in high-intensity exercise within 24 h before the test. The study protocol and ethics were approved by the Institutional Review Board of Beijing Sport University (2019018H).

2.2 Procedures

Before the gait test, the participant was in tight-fitting clothes and self-provided running shoes. Then participants warmed up and practiced to become familiar with the test process and ensure that they could step on the force plate naturally and accurately during the formal test. After that, a total of 23 reflective markers were placed on the participant's vertex of the head, the upper edge of the sternum, the midpoint of the fourth and fifth lumbar vertebrae, and bilaterally on acromions, anterior superior iliac spines, anterior thighs, medial femoral condyles, lateral femoral condyles, tibial trochanters, medial tibial condyles, lateral fibular condyles, heels, and toe tips. Surface

electrodes were placed bilaterally on the participant's long head of biceps femoris, semitendinosus, rectus femoris, vastus medialis, vastus lateralis, medial head of gastrocnemius, and lateral head of the gastrocnemius. The skin was shaved and exfoliated, then cleaned with alcohol before electrode placement.

In the gait test, participants performed repeated trials twice for each movement, including walking and jogging at their self-selected speed on both sides. During the test, participants started from 10 m away from the force plate and walked or jogged through the data collection area at their self-selected speed. During the whole process, participants kept smooth movements without pace adjustment. The action cycle of walking was defined from the tested foot strike and untested foot off to the tested foot strike again. The action cycle of jogging was defined from the tested foot strike to the tested foot strike again.

Muscle maximum voluntary contractions (MVCs) tests were performed after the gait test. During MVC tests of the medial head of gastrocnemius and lateral head of gastrocnemius, the participants were held in a standing position with their shoulders immobilized and used maximal effort to raise their heels. During MVC tests of the rectus femoris, vastus medialis, and vastus lateralis, participants were in a seated position with knee flexion angles of about 90 degrees. After fixing the participants' shank and thigh, they used maximal effort to extend the knee. During MVC tests of the long head of biceps femoris and semitendinosus, participants were in a seated position with knee flexion angles of about 90 degrees. After fixing the participants' shank and thigh, they used maximal effort to bend the knee. MVCs tests were repeated three times (5 s each) on both sides for each participant, with 1-min rest between repetitions.

2.3 Data collection

The reflective marker trajectories in the gait test were captured using a 14-camera motion capture system (Qualisys, Gothenburg, Sweden) at 200 Hz. The ground reaction force (GRF) in the gait test was obtained using three 3-D force plates (Kistler Instruments, Winterthur, Switzerland) at 1000 Hz. Electromyographic (EMG) data were recorded using a wireless test system (Delsys Trigno Mobile, USA) at 2000 Hz. The acquisition of marker trajectories, GRF, and EMG data were synchronized by signals from Qualisys external triggers.

2.4 Data processing and analysis

Marker trajectories, GRF, and EMG data were processed by an open-source Matlab (The Mathworks, MA) toolbox (MOtoNMS v2.2). GRF data were filtered using a second-order Butterworth low-pass filter at 50 Hz. The marker trajectories data were filtered using a second-order Butterworth low-pass filter at 13 Hz. The raw EMG data were removed mean, bandpass-filtered (second-order Butterworth, 50–450 Hz), full-wave rectified, and low-pass filtered (second-order Butterworth, 6 Hz) to produce linear envelopes. EMG envelopes were standardized using the maximum value during gait and MVCs tests.

To obtain a participant's personalized model, a generic musculoskeletal model (gait 2,392) was scaled in OpenSim version 4.0 (Stanford, CA, United States). The scaling setting detail was in the Supplementary Material (Supplementary Table S1). The 3-D marker trajectories and GRFs collected in the gait test were used to calculate joint angles and moments through the Inverse Kinematics and Inverse Dynamics (ID) tools in OpenSim. Muscle analysis tools in OpenSim were used to calculate muscle-tendon unit (MTU) lengths and moment arms.

Muscle forces were estimated using an EMG-assisted method in the OpenSim plug-in CEINMS (Pizzolato et al., 2015). Based on the purpose of this study, we only analyzed the muscle forces around knee joints. There were two stages for muscle force estimation. In the first stage, data from one of the two repeated trials were used to calibrate the neuromuscular parameters of the personalized model. Seven experimental muscle excitations were mapped to 12 MTUs based on previous studies (Lloyd and Besier, 2003; Sartori et al., 2014; Hoang et al., 2019) (Supplementary Table S2). Then, the calibration process was performed in CEINMS to adjust the neuromuscular parameters to minimize joint moment prediction errors for the knee flexion-extension. The neuromuscular parameters included tendon slack length, optimal fiber length, maximum isometric force, non-linear shape factor, and EMG-to-activation recursive filter coefficients (Lloyd and Besier, 2003). The default value of neuromuscular parameters was from the scaled model. In the second stage, the calibrated neuromuscular model was used to predict muscle forces through the EMG-assisted model in CEINMS in the remaining trials. The objective function was:

$$F_{obj} = E_{trackMOM} + E_{trackEMG} + E_{sumEMG}$$

where $E_{trackMOM}$ is the sum of the squared differences between EMG-assisted predicted and experimental joint moments, $E_{trackEMG}$ is the sum of the squared differences between EMG-assisted predicted and experimental muscle excitations for seven MTUs recorded in the gait test, E_{sumEMG} is the sum of squared muscle excitations for all 12 MTUs. After predicting muscle forces, tibial contact forces (knee joint applied to tibia segment) were calculated using the Joint Reaction Analysis tools in OpenSim. The reference coordinate system was the tibia coordinate system. Results of muscle forces, joint moments, and tibial contact forces were normalized to body mass.

The muscle co-contraction index (CCI) was used to reflect muscle activation strategy (Hoang et al., 2019). We analyzed the CCI of knee flexors-extensors ($CCI_{flex\text{-}ext}$) in this study. The calculating function was:

$$CCI_{flex-ext} = \begin{cases} 1 - \dfrac{M(ext)}{M(flex)}, & M(ext) < M(flex) \\ \dfrac{M(flex)}{M(ext)} - 1, & otherwise \end{cases}$$

where M (ext) is the mean value of knee extensors activation (rectus femoris, vastus medialis, and vastus lateralis, vastus intermedius); M (flex) is the mean value of major knee flexors activation (long head of biceps femoris, semitendinosus, semimembranosus, medial head of gastrocnemius, lateral head of the gastrocnemius). When $CCI_{flex\text{-}ext} = 0$, means full co-contraction; when $CCI_{flex\text{-}ext} = 1$, means no co-contraction and only flexor activation; when $CCI_{flex\text{-}ext} = -1$, means no co-contraction and only extensor activation.

Tibial compression forces calculated by the musculoskeletal were qualitatively compared with available instrumented knee

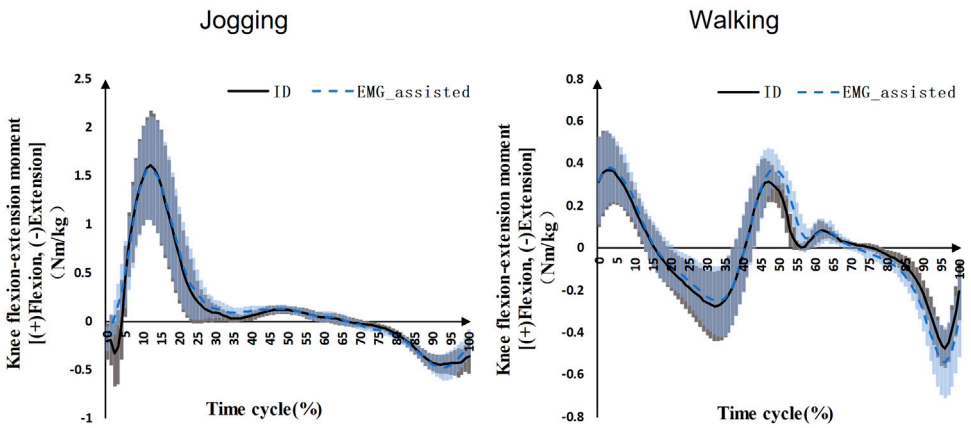

FIGURE 1
Knee flexion-extension net moments during jogging and walking (Black solid line: mean value of ID results on the both sides for seven patients, blue dashed line: mean value of EMG-assisted results on the both sides for seven patients, grey area: standard deviation of ID results on the both sides for seven patients, blue area: standard deviation of EMG-assisted results on the both sides for seven patients).

implant data in the OrthoLoad dataset (Bergmann and Graichen, 2008; Bergmann et al., 2014) (Jog: a 70-year-old male, weighing 81.5 kg and standing 1.74 m, who jogged at a constant speed of 1.67 m/s on a treadmill; Walk: a 74-year-old female weighing 69.1 kg and standing 1.66 m, who walked at a speed of 1.11 m/s on a treadmill with bare feet) (Supplementary Figure S1). The coefficient of determination (R^2) and root-mean-square error (RMSE) was calculated for the EMG-assisted predicted and ID knee flexion-extension net moment, which was used to evaluate EMG-assisted prediction results. Shapiro-Wilk test was used to determine the normality of lower limb biomechanical indexes on both sides. The paired sample *t*-test was used to analyze the difference in lower limb biomechanics between the participants' healthy and surgical sides. A type I error rate no greater than 0.05 was chosen as an indication of statistical significance. All statistical analyses were performed using SPSS 22.0 software (SPSS, Chicago, IL, United States).

3 Results

The knee flexion-extension net moment predicted by the EMG-assisted method showed high R^2 and low RMSE with ID results during jogging (R^2 = 0.95 ± 0.05, RMSE = 0.11 ± 0.05 Nm/kg) and walking (R^2 = 0.94 ± 0.04, RMSE = 0.06 ± 0.03 Nm/kg) (Figure 1).

Participants' action cycle on the healthy side was significantly longer than the surgical side during jogging (healthy = 0.741 ± 0.031 s, surgical = 0.721 ± 0.033 s, p = 0.009) and walking (healthy = 0.925 ± 0.023 s, surgical = 0.909 ± 0.026 s, p = 0.014) (Table 1). Participants' stance phase in jogging (healthy = 0.256 ± 0.013 s, surgical = 0.234 ± 0.017 s, p = 0.005) and swing phase in walking (healthy = 0.256 ± 0.013 s, surgical = 0.234 ± 0.017 s, p = 0.005) on the healthy side was significantly longer than the surgical side (Table 1). There was no difference in other characteristics.

Participants' peak tibial compression force on the healthy side (68.76 ± 11.95 N/kg) was significantly higher than the surgical side (61.86 ± 5.46 N/kg) during jogging (p = 0.039) (Figure 2). There was no significant difference in the first (healthy = 43.00 ± 12.74 N/kg, surgical = 35.05 ± 12.61 N/kg, p = 0.122) and second (healthy = 31.12 ± 8.96 N/kg, surgical = 31.72 ± 14.32 N/kg, p = 0.445) peak tibial compression forces during walking between the healthy and surgical side (Figure 2).

Participants' rectus femoris (healthy = 32.08 ± 10.79 N/kg, surgical = 22.58 ± 9.16 N/kg, p = 0.035) and vastus medialis (healthy = 6.08 ± 2.13 N/kg, surgical = 4.75 ± 2.18 N/kg, p = 0.036) forces on the healthy side were significantly larger than the surgical side at the peak moment of tibial compression forces during jogging (Table 2; Figure 3). There was no difference in other bilateral muscle forces at the peak moment of tibial compression forces during jogging and walking (Tables 2–4 and Figures 3, 4).

There was no difference in bilateral $CCI_{flex\text{-}ext}$ at the peak moment of tibial compression forces during jogging and walking (Table 5).

Participants' knee flexion angles (healthy = 49.50 ± 3.77°, surgical = 44.51 ± 7.03°, p = 0.042) and ankle dorsiflexion angles (healthy = 23.84 ± 6.50°, surgical = 19.88 ± 5.04°, p = 0.046) on the healthy side were significantly higher than that on the surgical side at the peak moment of tibial compression forces during jogging (Table 6). Participants' ankle dorsiflexion angles (healthy = 6.94 ± 3.85°, surgical = 4.91 ± 4.09°, p = 0.038) on the healthy side were significantly higher than that on the surgical side at the first peak moment of tibial compression forces during walking (Table 6). There was no difference in other bilateral joint angles at the peak moment of tibial compression forces during jogging and walking (Table 6).

4 Discussion

The purpose of this study was to compare the difference in bilateral tibial contact force for unilateral ACLR patients in order to assess the risk of knee osteoarthritis. The EMG-assisted method was used to predict muscle forces based on the neuromusculoskeletal model. We evaluated model simulation results in two ways. Firstly,

TABLE 1 Basic gait characteristics in jogging and walking.

	Jog	Jog	*p*-value	Walk	Walk	*p*-value
	Healthy side	Surgical side		Healthy side	Surgical side	
Step length (cm)	94.31 ± 5.53	95.76 ± 3.04	0.279	85.40 ± 3.09	82.72 ± 5.41	0.125
Velocity (m/s)	2.49 ± 0.13	2.54 ± 0.10	0.172	1.57 ± 0.06	1.62 ± 0.09	0.081
Action cycle (s)	0.741 ± 0.031	0.721 ± 0.033*	**0.009**	0.925 ± 0.023	0.909 ± 0.026*	**0.014**
Stance phase (s)	0.256 ± 0.013	0.234 ± 0.017*	**0.005**	0.485 ± 0.017	0.486 ± 0.024	0.411
Swing phase (s)	0.485 ± 0.040	0.487 ± 0.038	0.351	0.440 ± 0.010	0.422 ± 0.006*	**0.002**

*Significant difference between groups ($p \leq 0.05$).
The bold values were used to highlight the *p*-values ≤ 0.05.

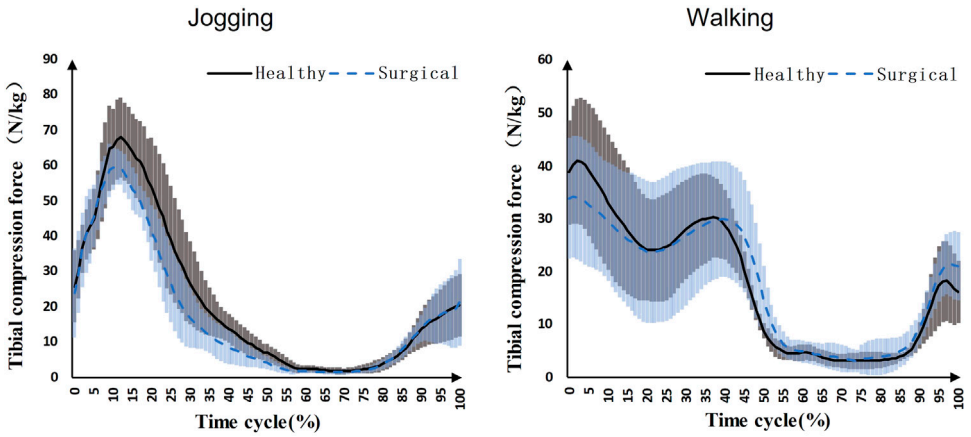

FIGURE 2
Knee compression forces on both sides during jogging and walking (Black solid line: mean value of results on the healthy sides for seven patients, blue dashed line: mean value of results on the surgical sides for seven patients, grey area: standard deviation of results on the healthy sides for seven patients, blue area: standard deviation of results on the surgical sides for seven patients).

TABLE 2 Muscle forces at the peak moment of tibial compression forces on the both sides during jogging.

Muscle name	Healthy side (N/kg)	Surgical side (N/kg)	*p*-value
rectus femoris	32.08 ± 10.79	22.58 ± 9.16*	**0.035**
vastus intermedius	6.56 ± 2.35	5.41 ± 2.29	0.122
vastus medialis	6.08 ± 2.13	4.75 ± 2.18*	**0.036**
vastus lateralis	9.10 ± 3.44	7.66 ± 3.29	0.073
long head of biceps femoris	1.53 ± 1.28	1.15 ± 0.46	0.390
semimembranosus	1.17 ± 0.73	0.96 ± 0.70	0.378
semitendinosus	0.63 ± 0.28	0.53 ± 0.25	0.395
short head of biceps femoris	1.53 ± 0.49	1.07 ± 0.52	0.095
lateral head of the gastrocnemius	2.73 ± 1.70	3.23 ± 0.95	0.453
medial head of gastrocnemius	8.22 ± 3.27	7.93 ± 1.15	0.775
gracilis	2.08 ± 1.22	1.60 ± 1.26	0.383
sartorius	0.13 ± 0.18	0.11 ± 0.10	0.676

*Significant difference between groups ($p \leq 0.05$).
The bold values were used to highlight the *p*-values ≤ 0.05.

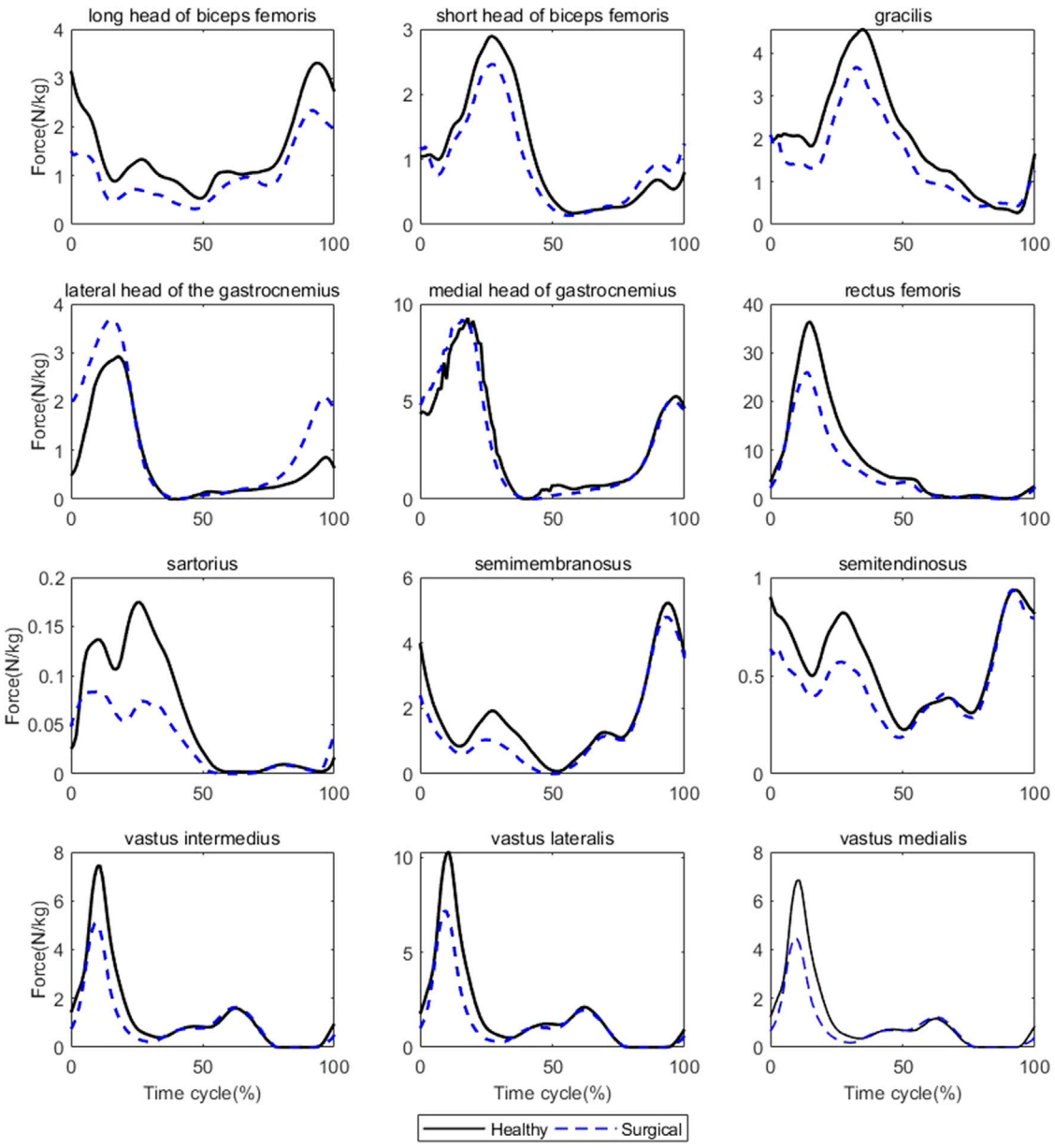

FIGURE 3
Muscle forces on both sides during jogging (Black solid line: mean value of results on the healthy sides for seven patients, blue dashed line: mean value of results on the surgical sides for seven patients).

the joint net moment calculated by EMG-assisted methods was compared to the ID joint net moment. Studies that use the EMG-driven method usually evaluated results in this way (Lloyd and Besier, 2003; Shao et al., 2009; Nikooyan et al., 2012; Sartori et al., 2012; Sartori et al., 2014). Our results showed a high correlation and low error between these two results. Secondly, the tibial compression force calculated by the model was compared with the measured values of knee implants obtained by the patients after total knee replacement when they performed similar movement tasks. Our results showed a similar trend and magnitude to Bergmann et al.'s results (Bergmann et al., 2014). Our results in jogging were larger than the Bergmann et al.'s results. This may be due to differences in the participants' age, jogging speed, and history of joint surgery.

The results of tibial compression forces support our first hypothesis that the tibial compression force on the surgical side for ACLR patients will be smaller than on the healthy side. This study showed that the peak tibial compression force on the healthy side was higher than on the surgical side during jogging. This result is consistent with the literature. One study followed a group of ACLR patients for up to 5 years after surgery and reported that those who developed osteoarthritis 5 years later had lower tibial compression forces in their gait 6 months to 2 years after surgery than those who did not (Wellsandt et al., 2016). Another study suggested that knee joints with ACLR surgery showed lower tibial contact forces during jogging, walking, and side stepping than the healthy group (Saxby et al., 2016).

TABLE 3 Muscle forces at the first peak moment of tibial compression forces on the both sides during walking.

Muscle name	Healthy side (N/kg)	Surgical side (N/kg)	*p*-value
rectus femoris	10.35 ± 3.98	8.17 ± 5.75	0.129
vastus intermedius	2.31 ± 1.33	1.76 ± 1.29	0.115
vastus medialis	2.00 ± 1.30	1.60 ± 1.45	0.198
vastus lateralis	3.26 ± 1.60	2.77 ± 2.12	0.215
long head of biceps femoris	1.44 ± 0.75	1.33 ± 0.92	0.814
semimembranosus	2.13 ± 0.99	2.34 ± 1.63	0.791
semitendinosus	0.74 ± 0.49	0.69 ± 0.53	0.855
short head of biceps femoris	2.18 ± 0.96	1.60 ± 0.74	0.171
lateral head of the gastrocnemius	1.89 ± 1.27	1.60 ± 1.73	0.534
medial head of gastrocnemius	6.58 ± 6.66	3.13 ± 2.61	0.221
gracilis	1.45 ± 1.46	0.81 ± 1.08	0.096
sartorius	0.16 ± 0.29	0.05 ± 0.05	0.275

TABLE 4 Muscle forces at the second peak moment of tibial compression forces on the both sides during walking.

Muscle name	Healthy side (N/kg)	Surgical side (N/kg)	*p*-value
rectus femoris	3.28 ± 2.02	5.00 ± 4.15	0.106
vastus intermedius	0.13 ± 0.12	0.23 ± 0.29	0.106
vastus medialis	0.11 ± 0.12	0.48 ± 0.90	0.138
vastus lateralis	0.16 ± 0.12	0.21 ± 0.22	0.130
long head of biceps femoris	0.33 ± 0.36	0.30 ± 0.46	0.630
semimembranosus	0.43 ± 0.38	0.51 ± 0.95	0.722
semitendinosus	0.22 ± 0.13	0.25 ± 0.41	0.765
short head of biceps femoris	0.88 ± 0.60	0.77 ± 0.73	0.560
lateral head of the gastrocnemius	3.19 ± 2.02	5.08 ± 4.36	0.302
medial head of gastrocnemius	9.88 ± 6.47	7.06 ± 4.39	0.187
gracilis	1.26 ± 1.10	1.24 ± 1.63	0.966
sartorius	0.02 ± 0.01	0.01 ± 0.02	0.635

Lower tibial compression force may lead to an increased risk of knee osteoarthritis. On the one hand, the thickness of healthy cartilage increases under a higher repetitive load and joint unloading is associated with regional cartilage thinning after ACLR (Koo and Andriacchi, 2007). A smaller tibial compression force may cause regional cartilage thinning and joint degeneration. The patient's physical activity level will gradually increase to return to daily life after surgery. If the joint load-bearing capacity has not recovered to its pre-operative level, the risk of knee osteoarthritis may be increased. On the other hand, Saxby et al. suggested that the total area of contact between the tibiofemoral articulating surfaces may reduce because of a smaller knee flexion angle and knee flexion excursion, so articular contact forces were focused to smaller regions

TABLE 5 $CCI_{flex\text{-}ext}$ at peak moments of tibial compression forces on the both sides during jogging and walking.

	Healthy side	Surgical side	*p*-value
Peak in jogging	−0.46 ± 0.32	0.30 ± 0.32	0.062
First peak in walking	0.24 ± 0.22	0.28 ± 0.19	0.296
Second peak in walking	0.86 ± 0.11	0.62 ± 0.40	0.051

although the magnitudes of tibial contact forces were small (Saxby et al., 2016). Our result also supported that the surgical side had a smaller knee flexion angle at the moment of peak tibial compression

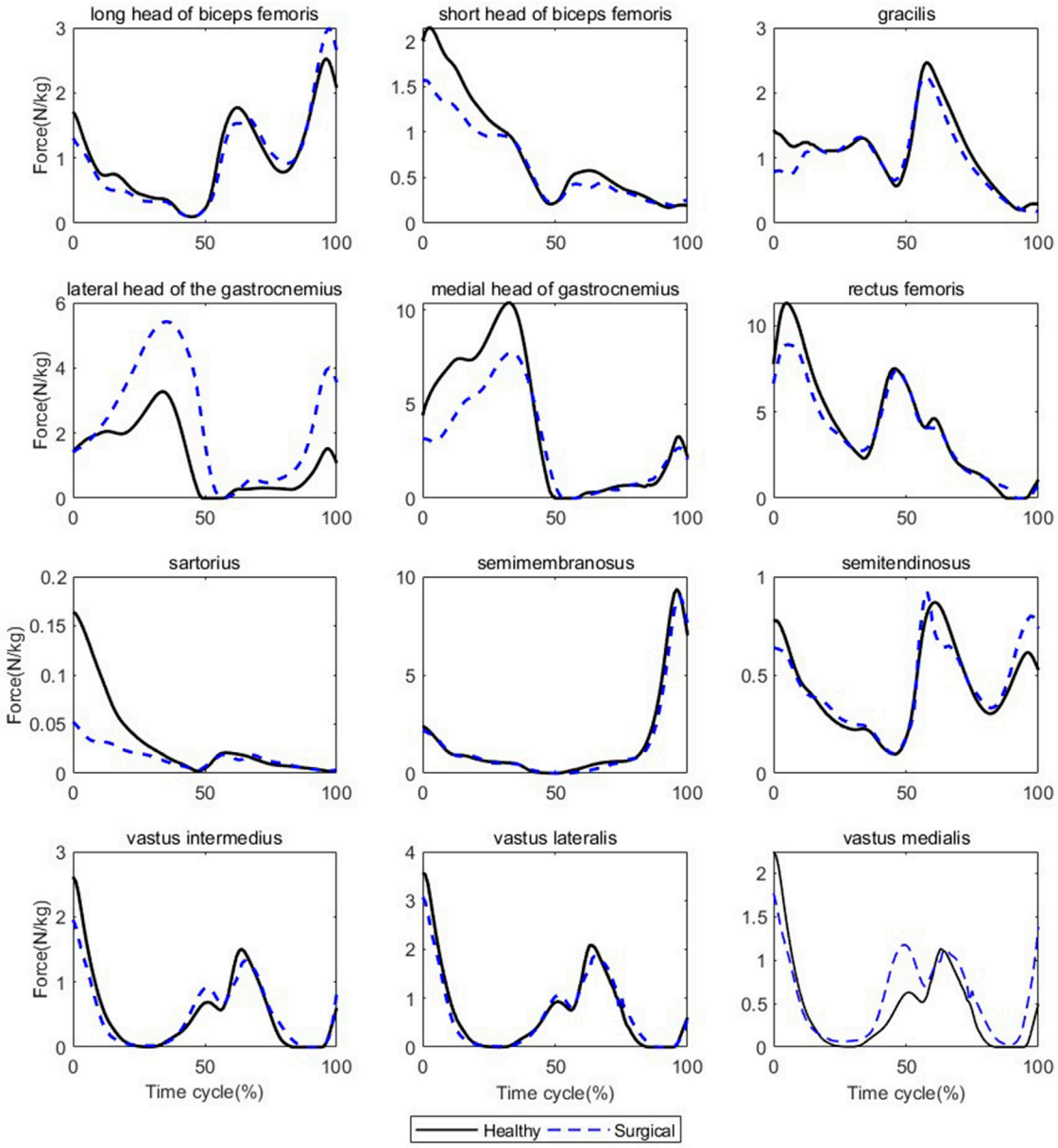

FIGURE 4
Muscle forces on both sides during walking (Black solid line: mean value of results on the healthy sides for seven patients, blue dashed line: mean value of results on the surgical sides for seven patients).

TABLE 6 Joint angles at peak moments of tibial compression forces on the both sides during jogging and walking.

	Peak in jogging			First peak in walking			Second peak in walking		
	Healthy side	Surgical side	*p*-value	Healthy side	Surgical side	*p*-value	Healthy side	Surgical side	*p*-value
ankle dorsiflexion (°)	23.84 ± 6.50	19.88 ± 5.04*	**0.046**	6.94 ± 3.85	4.91 ± 4.09*	**0.038**	19.41 ± 6.00	18.60 ± 3.96	0.574
knee flexion (°)	49.50 ± 3.77	44.51 ± 7.03*	**0.042**	30.12 ± 3.22	28.73 ± 4.87	0.241	11.89 ± 4.33	14.13 ± 4.16	0.089
hip flexion (°)	42.48 ± 11.49	41.56 ± 8.63	0.352	37.59 ± 9.95	40.84 ± 9.17	0.092	−4.41 ± 4.84	3.05 ± 6.59	0.573
hip adduction (°)	5.70 ± 3.41	5.36 ± 5.63	0.445	2.65 ± 5.06	0.34 ± 4.77	0.178	10.56 ± 1.88	8.60 ± 5.01	0.223
hip rotation (°)	7.17 ± 2.05	5.40 ± 3.02	0.120	12.46 ± 3.83	11.22 ± 4.14	0.327	5.92 ± 3.74	5.99 ± 6.06	0.951

*Significant difference between groups ($p \leq 0.05$).
The bold values were used to highlight the *p*-values ≤ 0.05.

force than the healthy side. This may be another explanation for the increased risk of knee osteoarthritis following ACLR.

The results of biomechanical characteristics support our second and third hypothesis that there will be differences in muscle forces around the knee joint and lower limb kinematics between the surgical and healthy sides during walking and jogging. The reason for the lower tibial compression force after ACLR may be the lower quadriceps muscle forces. We analyzed the lower extremity biomechanical characteristics at the moment of peak tibial compression force to understand why the peak tibial compression force on the surgical side is smaller than that on the healthy side. In this paper, it was suggested that compared with the healthy side, the ACLR side had a smaller knee flexion and ankle dorsiflexion angle at the peak moment of tibial compression force, which was consistent with the results of multiple previous studies (Lewek et al., 2002; Tsai et al., 2012; Kaur et al., 2016; Saxby et al., 2016). The reduced knee flexion angle is a typical change of gait after ACLR. However, this change does not appear to be the cause of the decreased tibial compression force since a smaller knee flexion and ankle dorsiflexion angle should theoretically increase the tibial compression force due to the direction of the force of gravity. Therefore, we infer that changes in knee and ankle joint motion patterns are not the primary reason for the decreased tibial compression force. Instead, muscle forces around the knee are directly related to the joint load (Saxby et al., 2016), and we found that the quadriceps femoris force on the surgical side was lower than that on the healthy side, which may be responsible for the reduced tibial compression force. Interestingly, we did not find any differences in $CCI_{flex\text{-}ext}$ between the two sides at the peak moment of tibial compression force. The activation patterns of muscles can affect joint contact forces, and that the high degree of co-contraction of muscles may increase joint compression forces. This inference is based on the positive correlation between muscle activation levels and muscle strength. However, although this study reported differences in bilateral tibial compression forces and muscle forces, it did not show any differences in $CCI_{flex\text{-}ext}$. One possible explanation is that the muscle model used in this study is based on the Hill model, which mainly includes the relationship between muscle force and muscle length, velocity, and activation level (Sartori et al., 2014; Pizzolato et al., 2015). $CCI_{flex\text{-}ext}$ only reflects the characteristics of muscle activation, but this is not the only factor that affects the muscle force.

The neuromusculoskeletal model for the EMG-assisted method in this paper only included knee flexion-extension degrees of freedom. Although the range of motion of the knee joint in jogging and walking is tiny, the influence of muscles around other joints on the development of knee osteoarthritis can be further analyzed in the future study. In addition, the sample size of this study was not large, and the participants' postoperative time ranged from 1.5 to 3.5 years, which does not fully explain the time frame for the increased risk of knee osteoarthritis, but additional participant data is being collected to test the approach across a larger cohort.

5 Conclusion

Unilateral ACLR patients had smaller tibial compression forces on the surgical side than on the healthy side while jogging, which may increase the risk of knee osteoarthritis. The main reason may be that the quadriceps muscles do not exert enough force. We recommend that ACLR patients pay attention to strengthening their quadriceps muscle through strength training during the postoperative rehabilitation process, and try to increase the use of the quadriceps muscle on the surgical side to an appropriate level in daily activities.

Data availability statement

The raw data supporting the conclusion of this article will be made available by the authors, without undue reservation.

Ethics statement

The studies involving human participants were reviewed and approved by the Sports Science Experiment Ethics Committee of Beijing Sport University. The patients/participants provided their written informed consent to participate in this study.

Author contributions

TL: Conceptualization, Methodology, Software, Formal Analysis, Writing Original Draft JF: Revising the manuscript critically for important intellectual content, Supervision HuL: Revising the manuscript critically for important intellectual content, Supervision HaL: Conceptualization, Funding Acquisition, Resources, Supervision.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Supplementary material

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fphys.2023.1160261/full#supplementary-material

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Methodological considerations for determining the volume and intensity of drop jump training. A systematic, critical and prepositive review

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This systematic review provides critical and propositional information on criteria for determining the volume and intensity of drop jumps during plyometric training programs. Eligibility criteria were defined according to PICOS: Participants: male or female athletes, trained or recreationally active (16–40 years). Intervention duration: more than 4 weeks. Comparators: passive or active control group during a plyometric training program. Outcomes: information on improvement with Drop Jump or Depth Jump, with other jumps, acceleration, sprint, strength, and power output. Design: randomized controlled trials. We searched articles published in PubMed, SPORTDiscus, Web of Science, and Scopus. The search was conducted until 10 September 2022 for English-language articles only. The risk of bias was assessed using Grading of Recommendations, Assessment, Development and Evaluation (GRADE) for randomized controlled studies. We identified 31495 studies, of which only 22 were included. We found that six groups presented results with women, 15 presented results with men, and the remaining four presented mixed studies. Of the 686 people recruited, 329 participants aged 25.79 ± 4.76 years were involved in training. Methodological problems in training intensity, volume distribution, and individualization were noted, but methodological recommendations for their solution are also provided. It is concluded that the drop height should not be understood as the intensity determinant of plyometric training. Intensity is determined by ground reaction forces, power output, and jump height, among other factors. Furthermore, the athletes' experience level selection should be based on the formulas recommended within this research. These results could be helpful for those who intend to conduct new plyometric training programs and research.

KEYWORDS

drop jump, depth jump, jumping program, plyometric training, plyometrics

Introduction

Plyometric exercises have been used for research and training programs since the 1960s. This training method has grown in popularity over the last 12 years. A PubMed search for the term "plyometric" yields 237 research articles published between 1983 and 2010, while 301 were published between 2021 and 2022 alone, highlighting the importance of the topic.

Among the most classic exercises are the Depth Jump (DeJ) and the Drop Jump (DJ), where the athlete stands upright on a box, and keeps his hands on his hips, determines the starting leg, and then performs a free fall in order to provoke a high force against the ground and take off in the shortest possible time in the case of the DJ, but reach a greater jump height (JH) in the case of the DeJ (Clutch et al., 1983) The differences between these tasks are presented in appendix 1. During the free fall, the neuromuscular system prepares to counteract the effects of gravity, a phase in which neuromuscular pre-activation (Taube et al., 2012; Di Giminiani et al., 2020) and proprioceptive receptors which are integrated into the central nervous system (Taube et al., 2012) intervene to ensure an effective landing. The athlete then comes into contact with the ground, where the stretch-shortening cycle (SSC) takes place (Dietz et al., 1979; Schmidtbleicher and Haralambie, 1981; Horita et al., 2003; Turner and Jeffreys, 2010; Pedley et al., 2022). This consists of an eccentric phase where the muscle-tendon unit of the already pre-activated contractile component (Bosco et al., 1982b; Taube et al., 2012) starts a braking phase and lengthens, causing an accumulation of elastic energy at the tendons (Ishikawa et al., 2006) and the production of mechanical impulse. This is followed by stabilization time (Wikstrom et al., 2004; Flanagan et al., 2008) and a transition from eccentric to concentric, where velocity = 0 is reached, after which the concentric phase of the movement begins, where the elastic energy is released as a result of the shortening of the muscle-tendon unit (Ishikawa et al., 2006), summating to the myotatic stretch reflex for a more powerful concentric action (Dietz et al., 1979).

The science fails to identify the first to introduce the term "plyometric," and it is shared between Margaria. R and Wilt. F (Wilt, 1976), The term plyometric is thought to be derived from the Greek "plio," which is associated with longer or wider and, "metric," which means to measure. As far as we reviewed, the earliest research reported in the scientific literature on plyometric exercise is provided by Asmussen and Bonde-Petersen (1974) and by Verkhoshansky and Chernousov (1974). Asmussen and Bonde-Petersen (1974) found that there was a 5% improvement in JH comparing the Squat Jump (SJ) to the countermovement Jump (CMJ) and 11% comparing the SJ with the presently known 40 cm DJ, demonstrating that a negative work phase preceding the jump, significantly increases jump height up to a point, as previously thought. Y. Verkhoshansky himself (Verkhoshansky, 2006) assumes that this concept of plyometric work had previously been characterized by Sechenov and Mikhailovich (1863), when he described the spring function of the muscle. The former, in his work with jumping athletes, exploited the kinetic energy accumulated in the legs as a result of free falls and emphasized that the plyometric method is not a simple deformation of the muscle as a result of stretching, but that this stretching has to be fast (Verkhoshansky, 2006). According to this criterion, he proposed a range of fall heights (FH) for the training of the lower limbs (0.10–2.2 m), and the concept of "optimal plyometric load" (0.75 m) based on the average power output, which has been strongly criticized to this day (Bobbert et al., 1987a; Bobbert et al., 1987b; Baechle et al., 2008). During the period 1976–1982, considerable contributions were made by C. Bosco, who, for instance, found that athletes improved their CMJ by 11% and their DJ by 15% by using DJ in their training program (Komi and Bosco, 1978b; Bosco and Komi, 1979a; 1980; Bosco and Komi, 1979b; Bosco et al., 1981; Bosco et al., 1982a; Bosco et al., 1982b; Bosco et al., 1982c). Bosco also compared the jumping ability of men and women (Komi and Bosco, 1978a). Together with Verkhoshansky's, Bosco's work on plyometric and ballistic exercises is considered seminal when it comes to of plyometric training (Bosco et al., 1982c; Bosco et al., 1982d; Viitasalo and Bosco, 1982; Bosco et al., 1983a; Bosco et al., 1983b; Bosco et al., 1986), nevertheless, plyometric training has continued to evolve over the years (Baechle et al., 2008; Morin et al., 2019).

During that time, other researchers also made contributions to the theory of plyometric training. Cavagna (Margaria et al., 1960; Cavagna et al., 1965; Cavagna and Citterio, 1974) was among the first to provide an experimental justification for plyometric training, conducting experiments on the negative phase of jumping contributing to the action of the contractile part of muscles. Hakkinen et al. (1985) were also among the first to compare electromyography characteristics following a 24-week training program, finding significant improvements in jumping performance. Bobbert. M, made essential contributions to the technical structure of the DJ, finding biomechanical differences, in his first study (Bobbert et al., 1987a), between the DJ with rebounding and the DJ with countermovement, and recommending the former for athletes seeking to improve the mechanical output of the knee extensors and plantar flexors. Nevertheless, his second study (Bobbert et al., 1987b) suggested a FH at 20 or 40 cm when investigating the training effects of rebounding DJ. Bobbert was then the first to scientifically justify a range of FH. (Lees and Fahmi, 1994b), A was the first to refute all previous FH criteria and contributions (Komi and Bosco, 1978b; Bobbert et al., 1987b), indicating that the best performance for a net increase in JH, instantaneous power, and other parameters occurred at 12 cm FH. There have been considerable contributions regarding the best conditions for power production (Pw) (Bosco et al., 1983b; Matic et al., 2015b; Morin et al., 2019), ground reaction forces (GRF), rate of force development (RFD) (Jensen and Ebben, 2007; Ebben et al., 2008; Flanagan et al., 2008; Kossow and Ebben, 2018a), and the reactive strength index (RSI), as well as the differences between CMJ and DJ (Young, 1995; Young et al., 1995; Flanagan et al., 2008; Struzik et al., 2016; Montoro-Bombú et al., 2022), although not always consistent. The emergence of these variables has led to the utilization of different criteria for assuming plyometric training intensity. However, all these parameters require an extensive use of laboratory equipment and analysis, which makes them difficult to implement on a daily basis for plyometric training programs.

In recent years, numerous studies have investigated the effects of plyometric training programs, including: injury prevention and obesity reduction (Nobre et al., 2017; van de Hoef et al., 2019; Zubac et al., 2019; Lee et al., 2020), lower limb power and optimal

TABLE 1 Inclusion and exclusion criteria.

Guidelines	Inclusion Criteria	Exclusion Criteria
Participants	Athletes between the ages of 16 and 40. Of either sex. Experienced and inexperienced subjects in plyometric training.	People under 16, over 40 and older adults.
Interventions.	It incorporated a PJT program with DJ with rebounding or countermovement. DJ with free arms. Isolated or combined. No overload. Paired control groups or experimental and control groups. Including pretest and posttest.	Rehabilitation programs, horizontal Drop Jump exercise, programs with supplementation (creatine, caffeine, or other), groups that performed DJ with overload or did not report the drop height used during the training program, DJ on unstable surfaces.
Comparisons	Passive or active control group comparisons during a plyometric training program.	Lack of comparisons, comparisons with coordination only, comparisons with change of direction only.
Results	Measurements of vertical jumps, jump height, sprint, acceleration.	Absence of results, not clearly describing the results.
Study designs and article type	Randomized controlled trials incorporating PJT both parallel, crossover, and cluster with more than 4 weeks and no limitation of completion. Original, peer-reviewed research in the English language.	Non-randomized, case reports, cross-sectional, retrospective studies, observational studies. conference abstracts, books and book chapters, published in non-peer-reviewed journals, websites, and preprint editions, non-human studies.

PJT = jump training program; DJ = drop jump.

TABLE 2 Search strategy for each database.

Date of the search	8/09/2022	8/09/2022	9/09/2022	10/09/2022
Date base	SPORTDiscus (EBSCOhost)	PubMed/ MEDLINE	WOS	Scopus
Applied database fields used during the search	Title, Abstract,	Title, Abstract	Topic (Title, abstract, author keywords, and Keywords Plus)	Title, Abstract, keyword
Restrictions for the search	None			
Examples of the strategy PubMed/ MEDLINE	("Plyometric"[Title/Abstract] OR "Depth Jump" [Title/Abstract] OR "Drop jump"[Title/Abstract] OR "Vertical jump"[Title/Abstract] OR "Jump*"[Title/Abstract] OR "Stretch shortening cycle"[Title/Abstract] OR "Stretch shortening cycle"[Title/Abstract]) AND ("Training"[Title/Abstract] OR "Training program"[Title/Abstract] OR "Training effect*"[Title/Abstract] OR "Training intervention"[Title/Abstract])			
Examples of the strategy SPORTDiscus	TI ("Plyometric" OR "Depth Jump" OR "Drop Jump" OR "Vertical Jump" OR "Jump*" OR "Stretch-shortening cycle" OR "Stretch shortening cycle") AND TI ("Training" OR "Training Program" OR "Training Effect*" OR "Training intervention")			
	AB ("Plyometric" OR "Depth Jump" OR "Drop Jump" OR "Vertical Jump" OR "Jump*" OR "Stretch-shortening cycle" OR "Stretch shortening cycle") AND AB ("Training" OR "Training Program" OR "Training Effect*" OR "Training intervention")			
Examples of the strategy Web of Science	"Plyometric" OR "Depth Jump" "Drop Jump" OR "Vertical Jump" OR "Jump*" OR "Stretch-shortening cycle" OR "Stretch shortening cycle" (Topic) AND "Training" OR "Training Program" OR "Training Effect*" OR "Training intervention" (Topic)			
Examples of the strategy Scopus	(TITLE-ABS-KEY ("Plyometric" OR "Depth Jump" OR "Drop Jump" OR "Vertical Jump" OR "Jump*" OR "stretch-shortening cycle" OR "Stretch shortening cycle") AND TITLE-ABS-KEY ("Training" OR "Training Program" OR "Training Effect*" OR "Training intervention"))			

loading (Miller et al., 2002; Young, 2006; Baechle et al., 2008; Terzis et al., 2016), running economy and, overall performance in dynamic activities (Besier et al., 2001; Balsalobre-Fernandez et al., 2016). This required the development of an effective training program where, we infer, that it was necessary to understand the different components of training, such as exercise volume, frequency, program duration, progression, and intensity. However, a review article (Ramirez-Campillo et al., 2020b) showed that 42% of the studies did not report the intensity of their plyometric training, probably due to methodological difficulties. Furthermore, published research showed no consensus on the best method for determining plyometric exercise intensity (Ebben, 2007; Ebben et al., 2008; Jarvis et al., 2016; Ramirez-Campillo et al., 2020b). On the other hand, although the benefits of plyometrics are well researched, we currently do not find a critical and solid report on the most used variables for the implementation of volume and intensity during such training programs. Therefore, the criteria to determine the main components of plyometric training are an aspect that requires further exploration. Considering the lack of consensus on this problem, it seems justified to systematize the existing criteria for the volume and intensity of plyometric training in the scientific literature. This knowledge may allow more precise monitoring of plyometric training load and provide valuable indications for professionals. Therefore, the purpose of this systematic review was to collate and evaluate the criteria for determining the volume and intensity of drop jumps in the available literature.

Methods

The study was conducted following the recently updated Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) guidelines. (Page et al., 2021; Page et al., 2022).

Eligibility criteria

Inclusion and exclusion criteria were established *a priori*. Table 1 shows the rigorously determined definitions for type of participants, intervention, type of comparisons, quality of outcomes and study design (PICOS). Randomized studies used a parallel or crossover group design (Spieth et al., 2016; Hariton and Locascio, 2018). These criteria were rigorously followed to reduce the risk of between-group bias and avoid systematic differences in confounding factors (Spieth et al., 2016; Hariton and Locascio, 2018). Therefore, the purpose of this systematic review was

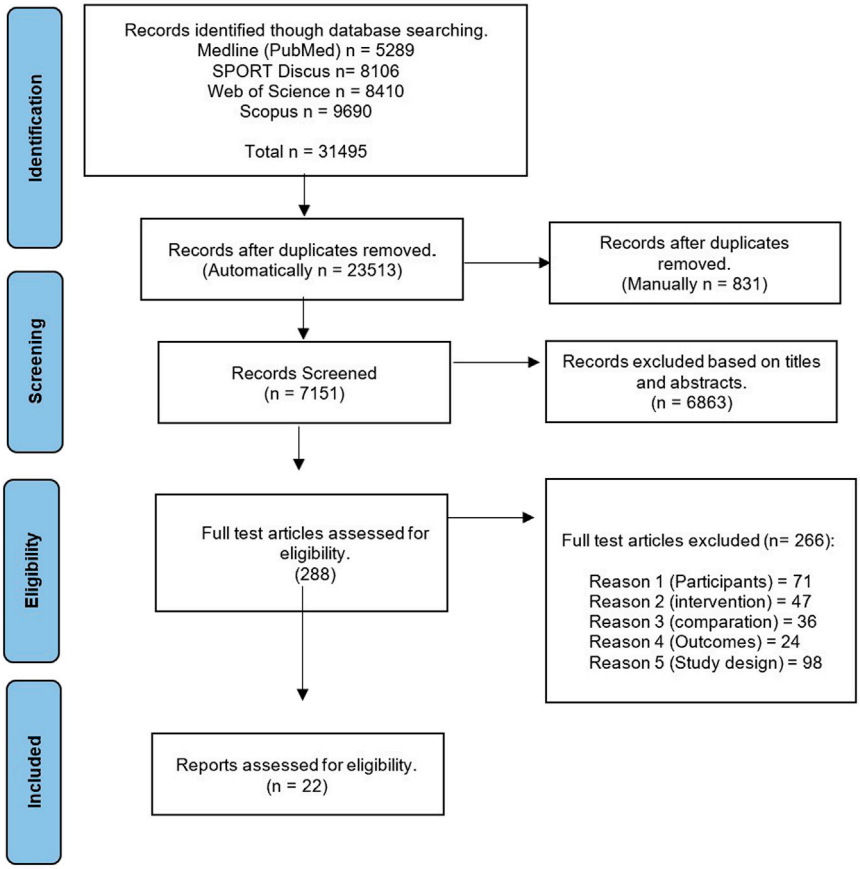

FIGURE 1
Prism flow chart on the results of the literature search.

to collate and evaluate the criteria for determining the volume and intensity of drop jumps in the available literature.

Sources information

The searches were carried out from the beginning of the indexing of the databases until the 10th of September 2022. The databases searched were PubMed, SPORTDiscus, Web of Science and, Scopus. Regulatory database records were used. We also contacted experts who provided papers that were not included and met the inclusion criteria. Reference searching of study citations was used as a source of information to detect potentially eligible studies.

Search strategy

Keyword selection was determined by experts and included: ("Training with adults" OR "Training with athletes" OR "Training with youth" OR "Program with adults" OR "Program with athletes" OR "Program with youth" OR "Effect*") AND ("Vertical Jump" OR "Plyometric*" OR "Drop Jump" OR "Depth jump"). Accounts were created in each database, automatically generating emails for information on new papers. These were received as available and were subject to the review process until the end of the study on the 23rd of December 2022. The search strategies are listed in Table 2.

Selection process

Two authors (RM and CB) developed the search process, removing duplicates using reference management software (EndNote TM X9, Clarivate Analytics, Philadelphia, PA, United States). They then performed title and abstract review and full-text analysis independently. By agreement between the two, studies in doubt of acceptance were separated for discussion, which was discussed and agreed upon with 100% agreement.

Data collection process and data items

The authors developed the data using criteria from other reviews (Ramirez-Campillo et al., 2018a; Ramirez-Campillo et al., 2020a). They were defined *a priori* and included the following information: 1 = name of authors and study; 2 = sport or activity; 3 = number of subjects, age (years), and sex; 4 = if the methodology states a plyometric training program; 5 = if the methodology states heights criteria before performing the exercise; 6 = if the program includes other jumps; 7 = if the methodology states the heights selected to execute the drop jump;

TABLE 3 Application of the GRADE scale for the RCTs included in the study.

	Quality assessment				Results summary			
					n-subjects		Absolute risk	
Studies	1	2	3	4	5	6	7	8
(Lum et al., 2022)	No pause between rep. No evaluation of Int. No volume variation criterion	M	JH	D	n=22	n=11	d=0.48	Mod.
(Falch et al., 2022)	No pause between rep. Insufficient volume No volume variation criterion	P	RSI	D	n=21	n=10	d=0.68	Bajo
(Sanchez-Sixto et al., 2021)	No assessment of Int. No volume variation criterion Does not report variation in reported heights.	P	JH	D	n=36	n=11	d=0.5	Mod.
(do Carmo et al., 2021)	No assessment of Int. No volume variation criterion	P	JH DJ	P	n=28	n=15	d=0.56	Mod
(Ahmadi et al., 2021)	No assessment of Int. No volume variation criterion	D	RSI JH	D	n=17 n=17	n=9 n=9	d=0.43 d=0.75	Mod.
(Laurent et al., 2020)	No assessment of Int. No volume variation criterion	D	JH	D	n=32	n=32	d=0.75	Mod
(di Cagno et al., 2020)	No assessment of Int. No volume variation criterion	D	JH CMJ	P	n= 54	n=28	d=0.27	Bajo
(Jeffreys et al., 2019)	No volume variation criterion	D	RSI 45cm	D	n=36	n=10	d=0.27	Mod
(Bogdanis et al., 2019)	No assessment of Int. No volume variation criterion	D	RSI	D	n=15	n=8	d=0.12	Mod
(Bianchi et al., 2019)	No assessment of Int. No volume variation criterion It does not report on the variation of reported heights.	P	(sprint 30 m)	P	n=21	n=11	d=0.4	Bajo
(Ramirez-Campillo et al., 2018d)	It does not report on the variation of reported heights.	N	JH	N	n=23	n=16	d=0.57	Alto
(Terzis et al., 2016)	No assessment of Int.	D	JH CMJ	P	n=20	n=10	d=0.66	Mod

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TABLE 3 (*Continued*) Application of the GRADE scale for the RCTs included in the study.

	Quality assessment				Results summary			
					n-subjects		Absolute risk	
(Ramírez-Campillo et al., 2014)	No assessment of Int.	D	JH CMJ	P	n=36	n=36	d=0.48	Mod
(Chelly et al., 2014)	No assessment of Int.	D	JH CMJ	D	n=23	n=12	d=1.00	Mod
(Ramírez-Campillo et al., 2013)	No assessment of Int.	D	JH CMJ	P	n=29	n=8	N/M	Mod
(Khlifa et al., 2010)	No assessment of Int.	D	JH CMJ	P	n=27	n=9	d=2.60	Mod
	No volume variation criterion							
(Sedano et al., 2009)	No assessment of Int.	D	JH DJ	D	n=20	n=10	d=0.77	Mod
	No volume variation criterion							
(de Villarreal et al., 2008)	No assessment of Int.	D	JH DJ 40	N	n=42	n=10	d=---	Mod
(Markovic et al., 2007)	No assessment of Int.	D	JH DJ 30	N	n=93	n=30	d= 0.90	Mod
(Luebbers et al., 2003)	No assessment of Int.	D	JH CMJ	P	n=38	n=19	d=0.37	Bajo
(Gehri et al., 1998)	No assessment of Int.	D	JH CMJ	D	n=28	n=11	d=0.22	Mod
	No volume variation criterion							
(Chelly et al., 2010)	No assessment of Int.	D	JH CMJ	D	n=23	n=12	d=0.33	Mod

1 = Program limitations; 2 = Program inconsistencies; 3 = Criteria for improvement; 4 = Bias; 5 = Total number of subjects; 6 = Total number of subjects intervened with DJ; 7 = Effect size found; 8 = GRADE quality; *working sessions, not weeks. **after four weeks of work. D = Disappeared; P = Present and N = Not present.

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TABLE 4 The objectives and main results of the included studies.

Studies	Objective	Main Results
(Lum et al., 2022)	To compare: Dynamic versus isometric strength- effect on JH.	Both plyometric and isometric exercises improved JH.
(Falch et al., 2022)	To investigate: Strength vs plyometric exercises on change of direction.	Both groups improved, but strength training was effective, and plyometric training was not sufficient.
(Sanchez-Sixto et al., 2021)	To compare: PJT vs combined PJT on the kinetics and kinematics of the vertical jump.	Combined PJT could provide better results in jumping performance than PJT alone.
(do Carmo et al., 2021)	To evaluate: PJT vs stimulation To verify: PJT vs perceived exertion, running pace and affective feeling.	PJT shows no change in stimulation. PJT shows no change in rhythm, perceived effort, and affective feeling.
(Ahmadi et al., 2021)	To evaluate: PJT on sand vs hard surface on the biomechanical variable.	PJT on hard surface had better results for biomechanical variables than sand surface.
(Laurent et al., 2020)	To compare: PJT DJ with flexed and extended knees on Achilles tendon stiffness and JH.	PJT increased JH in both groups, but JH of DJ20 cm and tendon stiffness showed better results for the group with extended knees.
(di Cagno et al., 2020)	To evaluate: PJT using a device vs PJT on range of motion, explosive and reactive strength.	PJT for both groups show no improvement in lunge distance, explosive and reactive strength
(Jeffreys et al., 2019)	To identify: Effectiveness of high and low volume plyometric loads on the stretch-shortening cycle.	Low volume PJT elicited the same improvement in RSI performance as a high-volume program.
(Bogdanis et al., 2019)	To compare: Unilateral versus bilateral PJT on maximal strength and rate of force development.	Unilateral PJT shows better results in strength and rate of force development for each leg individually.
(Bianchi et al., 2019)	To compare: High and low volume PJT on jumping, sprinting and change of direction abilities.	Both high and low-volume PJT induce benefits in jumping, sprinting and change of direction tests.
(Ramirez-Campillo et al., 2018d)	To compare: PJT with one session versus two sessions per week at the same weekly volume on fitness components.	PJT with one or two sessions has no additional effects on fitness development when the jumping volume is matched.
(Terzis et al., 2016)	To investigate: Effects of low intensity running vs power training on power development.	PJT has better results in JH and maximal power. Maximal isometric strength, rate of force development and maximal strength.
(Ramírez-Campillo et al., 2014)	To examine: Short-term PJT on explosive strength and endurance.	PJT improved running time, CMJ and DJ performance.
(Chelly et al., 2014)	To determine: Substitution of short-term PJT for some existing exercises would improve explosive movements.	Bi-weekly PJT improves important components of handball performance.
(Ramírez-Campillo et al., 2013)	To examine: Short-term PJT with different volumes and surfaces on neuromuscular performance.	High-volume PJT improves explosive performance, and PJT on hard surfaces increases explosive performance.
(Khlifa et al., 2010)	To examine: loaded versus unloaded PJT on vertical jump.	PJT with weights presents better results on vertical jumps.
(Sedano et al., 2009)	To examine: PJT on explosive strength, kicking speed and body composition.	PTJ improves explosive strength and kicking speed.
(de Villarreal et al., 2008)	To examine: PJT vs different frequencies and volumes on maximal strength, vertical jump and running.	PJT frequency two moderate volume sessions produce similar improvements at a greater efficiency than high volume frequency 4. Also, similar results in strength, jumping and running.
(Markovic et al., 2007)	To evaluate: PJT versus speed training on muscle function and dynamic athletic performance.	Sprint training produces similar or even more significant effects on muscle function and dynamic performance.
(Luebbers et al., 2003)	To determine: PJT of 4 vs seven weeks, followed by four weeks without plyometric training, on vertical jump performance and anaerobic power.	The four- and seven-week plyometric programs are equally effective in improving vertical jump height and anaerobic power after a 4-week recovery period.
(Gehri et al., 1998)	To determine: PJT of different techniques versus vertical jump capacity, positive energy production and elastic energy utilization.	Neither technique improved elastic power. In activities involving stretch-shortening cycles, DJ training was superior to CMJ.
(Chelly et al., 2010)	To examine: Combined hurdle and depth jump PJT on maximal power, strength, JH and muscle volume.	Biweekly PJT improved components of athletic performance relative to standard in-season training.

8 = if it presents volume parameters by weeks or sessions; 9 = if the methodology declares criteria for volume variation; 10 = training frequency and a total number of sessions (days/week); 11 = recovery between exercises (seconds or minutes); 12 = recovery between sets (seconds or minutes); 13 = presents the type of work surface (asphalt, grass, and others.); 14 = if the study presents cross-training and a total number of sessions (days/week)); 15 = if the study presents crossover; and 16 = Increase in performance shown by the program (Eq. (1)). A data extraction sheet was developed and tested with ten randomly selected studies, adapted from the Cochrane Consumers and Communication Group's data extraction template (https://cccrg.cochrane.org/cochrane-author-resources).

TABLE 5 Characteristics of the RCTs included in the study.

Studies	1	2	3	4	5	6	7		8	9	10	11	12	13	14
							weeks	vol							
(Lum et al., 2022)	Athletics endurance	n=22 e= 37±6 s= mix	No	No	DJ +	40, 50 y 60 cm	1 2-5 6	3 x 5 4 x 5 2 x 5	No	6 weeks. 2 x W Total: 18	no	3	No	No	9.20 % (JH)
(Falch et al., 2022)	Handball	n=21 e= 17±2 s= Fem	No	RSI	DJ +	40 cm	1 2 -3 4 5 6 7	1 x 6 1 x 8 1 x 10 1 x 12 1 x 8 1 x 5	No	8 weeks. 1 o 2 x W Total: 12	no	>2	Contact mat	No	37.0 % (RSI)
(Sanchez-Sixto et al., 2021)	Basketball	n=36 e= 23±6 s= Mal	No	No	DJ	20, 25, 30 y 40 cm	1-3 4 5 6	3 x 6 3 x 4 4 x 4 3 x 3	No	6 weeks. 2 x W Total: 12	1	3	No	No	5.71 % (JH)
(do Carmo et al., 2021)	Recreational runners	n=28 e= 33,3 s= Mal	No	No	DeJ +	40 cm	4-5 6 7 8	2 x 6 3 x 6 4 x 6 2 x 6	No	8 weeks 2 x W Total: 16	-	-	No	No	7.28 % (JH DJ)
(Ahmadi et al., 2021)	Volleyball	n=17 e= 23±2 s= Fem	No	No	DeJ +	40 cm	1-8	2 x 6	No	8 weeks 2 x W Total: 16	5 to 15´´	30 to 60´´	Wood vs sand	No	13.6 % (JH) 11.5 % (RSI)
(Laurent et al., 2020)	Physically active	n=32 e= 22.5 s= Fem	No	No	DJ +	20, 40 y 60 cm	1 2 3-10	3 x 10 4 x 10 5 x 10	No	10 weeks 2 x W Total: 20	2´	1.5 to 3´	Ground	No	13.7 % (JH)
(di Cagno et al., 2020)	Fencing	n=54 e= 17.5 s= Mal	No	No	DJ	50 cm	1-2 3-4 5-6	1 x 7 2 x 7 3 x 7	No	6 weeks 2 x W Total: 12	2´		Ground	No	4 % (JH) CMJ
(Jeffreys et al., 2019)	Rugby	n=36 e= 20±1 s= Mal	No	RSI	DJ	individual	2-6	2 x 5	Does not vary	4 weeks 2 x W Total: 8	-	-	No	No	15.78 % (RSI)
(Bogdanis et al., 2019)	Moderately trained	n=15 e= 19.6 s= Mix	No	no	DJ +	30 cm	1-3 4-6	2 x 10 3 x 10	No	6 weeks 2 x W Total: 12	-	1	No	No	4.30 % (RSI)
(Bianchi et al., 2019)	Football	n=21 e= 17±8 s= Mal	No	No	DJ +	60 cm	1-8	4 x 5	Does not vary	8 weeks 2 x W Total: 16	-	-	No	No	1.43 % (Sprint 30 m)
(Ramirez-Campillo et al., 2018d)	Football	n=23 e= 21±3 s= Fem	No	No	DJ +	5-35cm	1-2 3-4 5-6 7-8-	5 x 16 5 x 20 5 x 24 5 x 28 5 x 14	No	8 weeks 2 x W Total: 16	5-15´´	30-60´´	Assorted grass and ground	No	
(Terzis et al., 2016)	Physical Education students	n=22 e= 20±2 s= Fem	No	No	DJ +	20, 30 y 40 cm	1-6	8 x 3	No	6 weeks 1 x W Total: 6	30´´	1´	Contact mat	No	10.90 % (JH CMJ)
(Ramírez-Campillo et al., 2014)	Athletics endurance	n=36 e= 22.1 s= Mix	No	No	DJ +	20, 40 y 60 cm	1-6	6 x 10	Does not vary	6 weeks 2 x W Total: 12	15´´	2´	Wood	No	8.86 % (JH CMJ)
(Chelly et al., 2014)	Handball	n=23 e= 17.2 s= Mal	No	No	DJ +	40 cm	5-8	4 x 10	Does not vary	9 weeks 2 x W Total: 18	5´´		No	No	9.52 % (JH CMJ)
(Ramírez-Campillo et al., 2013)	Physical Education students	n=29 e= 16.86 s= Mal	No	No	DJ	20, 40 y 60 cm	1-6	6 x 10 12 x 10	Does not vary	7 weeks 2 x W Total: 14	5´´	1,5´	Wood OR mattress (3 cm)	No	N/M (JH CMJ)
(Khlifa et al., 2010)	Basketball	n=27 e= 23.61 s= Mal	No	No	DJ +	40 cm	1 2 3-10	3 x 5 5 x 9 6 x15	No	10 weeks 3 x W Total: 30	15 a 30´´	-	No	No	7.0 % (JH CMJ)

(Continued on following page)

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TABLE 5 (*Continued*) Characteristics of the RCTs included in the study.

Studies	1	2	3	4	5	6	7		8	9	10	11	12	13	14
							weeks	vol							
(Sedano et al., 2009)	Football	n=20 e= 22.9 s= Fem	No	No	DJ +	50 cm	1-6 2-4 3-5 8-10 9-11 12	8 x 5 10 x 5 9 x 5 11 x 5 10 x 5 9 x 5	No	12 weeks 1 x W Total: 12	1′	4′	Synthetic hard floor	No	18.0 % (JH DJ)
(de Villarreal et al., 2008)	Physical Education students	n=42 e= 23.5 s= Mal	No	No	DJ	20, 40 y 60 cm	1-7	6 x 10	Does not vary	7 weeks 4 x W Total: 28	-	1′	3 cm mattress	No	18.78 % (JH JD)
(Markovic et al., 2007)	Physical Education students	n=93 e= 20.1 s= Mal	No	No	DJ +	40 cm	8-11	4 x 10	No	4 weeks 3 x W Total: 12	5′′	-	Athletics track	No	14.2 % (JH JD)
(Luebbers et al., 2003)	Physical Education students	n=38 e= University s= Mal	No	No	DeJ +	40 cm	3 4 5-6 7	2 x 10 3 x 10 6 x10 7 x 10	No	5 weeks 3 x W Total: 15	15-30′′	-	No	No	4.0 % (JH CMJ)
(Gehri et al., 1998)	Recreationally active	n=28 e= 20.3 s= Mix	No	No	DJ	40 cm	1-2 3-10	2 x 8 4 x 8	No	10 weeks 2 x W Total: 20	5′′	1′	No	No	0.69 % (JH CMJ)
(Chelly et al., 2010)	Football	n=23 e= 19±0.7 s= Mal	No	No	DJ +	40 cm	5-8	4 x 10	Does not vary	9 weeks 2 x W Total: 18	5′′		Grass	No	2.43 % (JH CMJ)

1 = Sport or activity; 2 = Number of subjects, age and gender; 3 = Methodology states a plyometric preparation program; 4 = Methodology states height criteria before performing the exercise; 5 = Program includes DJ, DeJ plus other jumps; 6 = Methodology states the height selected to perform the DJ or DeJ; 7 = Volume structure by weeks or sessions; 8 = Methodology states criteria for volume variation; 9 = Frequency of training and the total number of sessions; 10 = Recovery between exercises; 11 = Recovery between sets; 12 = Type of work surface; 13 = The study presents crossover; and 14 = Increase in performance shown by the Program.

$$Pi\,(\%) = \frac{(Cr - Pr)*100}{Pr} \quad (1)$$

Where: Pi (%) = Performance increase.
Pr = Previous result.
Cr = Current result.

Studies risk of bias assessment

The methodological quality of the chosen studies using the recommended language for sports science (Winter et al., 2016) was assessed using the Grading of Recommendations, Assessment, Development and Evaluation (GRADE) model (Aguayo-Albasini et al., 2014). This model is advocated for RCTs not only because of the strength of the recommendations, but because it is based on the quality of the studies, risk balance, etc. According to its recommendations, the quality of evidence is classified as high, moderate, low, and very low. To individualize the level of recommendation and methodological quality, a specific modification of this scale was made for this study. This consisted of reporting individual studies only, thus excluding group studies, on the same variable (Guyatt et al., 2011).

Result

A total of 31495 studies were identified in the databases and exported to the bibliographic reference management software (EndNote TM X9, Clarivate Analytics, Philadelphia, PA, United States). A total of 23513 studies were automatically removed as duplicates, and a further 831 studies were manually removed as duplicates. The remaining studies (7151) were screened by title and abstract, taking into account the inclusion and exclusion criteria and then their relevance, resulting in the elimination of further 6863 studies. A total of 288 studies were eligible for full-text review. Once the full-text review was completed, 266 studies were excluded based on the following criteria: participants (71), intervention (47), comparison (36), outcomes (14), and study design (98). Thus, a total of (22) were included in the review for critical review (Figure 1).

Quality assessment and risk of bias of the studies chosen for the review

According to the GRADE assessment (Table 3), the study quality was average, with only one study that we considered to be of high quality. Applying the intraclass correlation index, it was found that the independent reviewers involved in the eligibility had a reliability of 0.89.

Study characteristics

The RCT included in this review recruited groups of physical education students and recreationally active individuals (n = 8), moderately trained cross-country athletics groups (n = 4), national level Football groups (n = 4), national level Handball groups (n = 2), national level Basketball groups (n = 2). The remaining groups were

TABLE 6 Methodological consideration for the determination of participants' experience in plyometric and ballistic activity.

Variables/ points	5	4	3	2	1
A. Type of population	International participation in competitions	Participation in national competitions	Participation in regional competitions	Systematic recreational sports practice	Sedentary or occasional sporting activity
B. Type of sport	Maximum, explosive, and reactive strength	Team sports.	Combat sports	Endurance, artistic, and Water sports	Random practice of team or individual sports, or no practice
C. Time experience in plyometrics	≥ 6 years	5-4 years	3-2 years	≤ 1 year	Never
D. IRSI 40 cm	≥ 1.75	1.74–1.25	1.24-1.00	0.99-0.50	≤ 49
E. CMJ-JH	≥ 50 cm	49-42 cm	41-34 cm	33-21 cm	≤ 20
Level of plyometric experience	Level 4 = ≥20 Long and systematic experience of extensive and intensive plyometric exercise	Level 3 = 13–19 Experience in plyometric development	Level 2 = 7–12 Basic plyometric experience	Level 1 = 1–6 Low introductory experience	Level 0 = ≤ 1 No Experience
Equation	Participant's level of plyometric experience* = [(A+B+C/2) /3] x (D+E)*Individualized criteria				

IRSI = integral reactive strength index; JH = jump height; CMJ = countermovement jump.

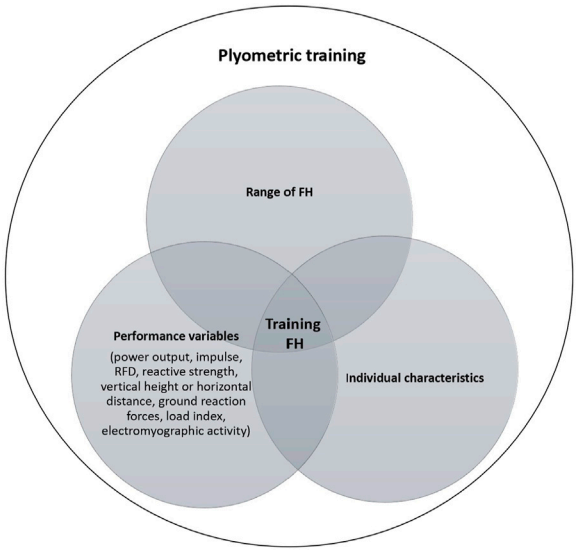

FIGURE 2
The determination process for the optimal training FH.

Rugby, Fencing, Volleyball and untrained subject. This amounted to 686 recruited individuals, of which n = 337 were intervened with DJ and DeJ training. The mean age of all participants was 25.79 ± 4.76 years, but it is worth noting that one study (Luebbers et al., 2003) did not present the age of the participants, only reporting that they were university students. Although it can be inferred that they were between 18 and 25 years old, the study was excluded from the calculation of the mean age. Of the 22 included RCT, six groups presented results with women only, 15 presented results with men only, and the remaining 4 presented a mixed study. The duration of plyometric training programs ranged from 4 to 12 weeks, while training frequencies were between 1 and 4 times per week. Twenty-one of the 22 included studies did not present a selection criterion for individual FH in the procedure. Twenty-one of the 22 included studies standardized the FH, with ranges between 20 and 60 cm. We also found that the total jump per session ranged between 10 and 140. However, all the 22 included studies did not present a procedure criterion for the selection of the plyometric work volume. Thus, 18 of the included studies present a variation of the training volume without methodological justification, and the remaining seven do not explain why they maintain the same work volume during all the weeks of training. The type of landing surface is not reported in 10 studies, and two others are considered unclear. The objectives and main results of the included studies are detailed in Table 4, while Table 5 shows the general characteristics of the studies in this systematic review.

Discussion

The purpose of this systematic review was to collate and evaluate the criteria for determining the volume and intensity of drop jumps within training programs, according to the available literature. The review protocol was registered on the International Platform of Registered Systematic Review and Meta-Analysis Protocols available at https://inplasy.com/inplasy-2021-2-0051/. So far, this is the first systematic review that analyze the literature regarding the volume and intensity of DJ and DeJ in plyometric training. In opposite to other scoping reviews (Ramirez-Campillo et al., 2020c) this is critical and propositional. Under the guidance of Boolean operators (AND and OR), this review found a total of 31495 studies in different databases, which were processed to finally include 22 studies in our review. The results reveal that, for plyometric training, there is still a lack of consensus in determining the volume and intensity of exercises. There is a need to deepen the quality of research processing in order to clearly determine volume and intensity of plyometric training, thus its effect. Therefore, in the present discussion, we aim to address the discordance within the

literature by developing operational frameworks that facilitate the understanding and design of training programs. Taking into account variables such as the athletes' experience level, along with training volume, intensity, density, duration, weekly frequency, exercise type and execution criteria, as well as a practical rationale for implementation.

Main criticisms and observations regarding training programs

Most of the selected studies showed a limited knowledge of the main characteristics of plyometric training. In fact, 90.90% (n = 20) of the reported study designs do not specify some variables that greatly affect the training effect (Bompa and Buzzichelli, 2019). For example, for what concern the landing surface, most studies do not report either the main objective of the plyometric work or the criteria of selection for the surface type. It is known that the surface type influences the stretch-shortening cycle (Bobbert et al., 1987a; Ramírez-Campillo et al., 2013) and that soft surface, such as grass or sand, do not allow contact times to be minimized, thus increasing reaction times. Hard surfaces on the other hand, allow shorter contact times, which guarantees a better reactivity (Ramírez-Campillo et al., 2013). These criteria are sometimes reported with a lack of clarity (de Villarreal et al., 2008; Sedano et al., 2009) or are contradictory to the stated objective (Chelly et al., 2010), but one study is quite clear with its statement related to the type of surface (Ramirez-Campillo et al., 2018d). Considering that the type of surface is closely related to the training effects of plyometrics, not stating the surface used in a study can lead to problems both in replicating it and in interpreting its results (Ramírez-Campillo et al., 2013; Ramirez-Campillo et al., 2020c). None of the included studies analyses the timing of utilization of hard and soft surfaces within the athletes' preparation phase. Although we can clearly infer the best time to use with different surfaces from the results of some studies (Ramírez-Campillo et al., 2013; Arazi et al., 2014; Ahmadi et al., 2021; Jacobs et al., 2021; Lannerstrom et al., 2021; Pereira et al., 2022). It would be interesting for future research to identify which type of surface can be the best choice according to the phases of the annual plan (general preparation, specific preparation, or competition phase).

Previous studies (Ramirez-Campillo et al., 2015), have indicated that plyometric training programs combined with other strength training means have better results than plyometric training alone. Although logical and very clear, this reasoning could be considered a tautology that needs to be reanalyzed. The combination of DJ training with strength training (Sanchez-Sixto et al., 2021), can present improvements in the kinetics and kinematics of the vertical jump, but this does not necessarily mean that 100% of the DJ training goals or 100% of the strength training goals have been met, and it would be implausible to determine the contribution of each one to the end result. The same way, when combinations of DJ, CMJ, SJ and multiple other jumps are performed (Table 5), although this is known to contribute to the corresponding training adaptations (Ramírez-Campillo et al., 2015; Ramirez-Campillo et al., 2015), researchers are uncertain to what extent each exercise affected the outcome. In this sense, while acknowledging the contributions of mixed plyometric studies, this research encourages the return to studies where the effects on performance variables of plyometric training, as a whole and in its components, can be isolated. Another fundamental element to be considered, is the population of such research studies. Several studies report implementing plyometric training programs with recreationally active individuals, physical education students or athletes with no previous experience in plyometric training (Table 5). In all these cases, care should be taken in interpreting the results: as in any activity performed with untrained subjects, if injuries do not occur, positive effects are most likely to be expected. Conversely, we can presume that when working with a population composed of experienced athletes, it may be difficult to obtain large effect sizes and large statistically significant differences due to the high level of specific work accumulated over the years.

An important variable that, in our opinion, is under-recognized, is the density of plyometric training. The included studies only refer sparingly to the weekly frequency of work, and none study indicates the density of the session. Density can be interpreted as the ratio between the total duration of a session, and the actual work performed (Bompa and Buzzichelli, 2019); for that reason, rest intervals become a determinant of the session density. Consequently, during plyometric training, the longer the recovery time, the lower the training density. This variable should also be considered when appraising training intensity, as it could be affected in the actual intensity in terms of power output (Lawton et al., 2006).

Another element that is too often overlooked, is the identification of the participants' level of experience. A previous report (Ramirez-Campillo et al., 2020c) had already shown that dichotomous variables (yes or no) do not gather sufficient information about athletes. We note that other studies (Sanchez-Sixto et al., 2021) report the athletes' years of training experience, but they do not specify whether the individuals came from the practice of team sports, power, or endurance sports, nor is it clear whether the experience (years) in plyometric training is systematic or just occasional. These criteria may be necessary to identify *a priori* whether large effect sizes can be expected at the end of the training interventions, as more experienced subjects need training means of higher specificity and intensity to deliver comparatively minimal, yet important, training effects. There is clearly an urgent need to standardize these elements.

As a consequence, although we recognize that experience is a multidimensional factor, we agree that there are general characteristics that mediate the influence of experience. Based on recommendations from previous reviews and meta-analyses (Moran et al., 2019; Ramirez-Campillo et al., 2020c; Moran et al., 2021; Ramirez-Campillo et al., 2021; Clemente et al., 2022), we propose five elements that we consider helpful in highlighting the participants' level of experience in plyometric and ballistic activities: (A) The type of sport population. This is related to the level of involvement in physical activity and sports. (B) The type of sport practiced. This is determined by the performance characteristics of the sport practiced and its relationship with plyometric actions. (C) Time of experience with either systematic or interspersed practice of plyometric exercise. (D). The level of the participant's integral reactive strength, and (E) the height reached in a CMJ. All these parameters were modified and adapted to the context following previous recommendations (Swann et al., 2015) and are shown in Table 6. Although it is beyond the scope of this

review to list all the possible sporting activities for each level, this table can be a crucial tool to standardize the criteria to assess and state the athletes' level of experience in new research.

Individualization of training is considered one of the fundamental principles of the sports training process (Bompa and Buzzichelli, 2019). However, individualization is not factored in by the researchers of most of the included studies. Only two studies (Ramirez-Campillo et al., 2018d; Jeffreys et al., 2019) reported individualizing the heights of fall, but still presented problems with the individualization of volume and intensity parameters. Likewise, the studies with equaled volume and FH (Table 5) present significant performance increases in the results but with great inter-individual outcome differences. For instance, 40 cm of height may represent 70% of the jumping capacity for one subject, but 90% for another. Similarly, a pre-set volume of, for example, 4 sets of 10 repetitions, may be optimal to improve a subject's power output while for another subject it could represent a power-endurance type of work, thus eliciting very different adaptations, not in line with the scope of the intervention. Although one study (Ramirez-Campillo et al., 2018c) considers plyometric individualization impractical for team sports, we considered it an essential element when working with high-performance sport.

Critical discussion on the use of the fall height as a determinant of intensity

Several authors emphasize that the FH is the sole determinant of the DJ and the DeJ' intensity. Yet, FH alone does not generate intensity and needs to be related to a performance variable associated with the exercise. Based on this systematic review, we understand that plyometric training intensity is determined by the association of a quantifiable variable (e.g., power output, impulse, RFD, reactive strength, vertical height or horizontal distance, ground reaction forces, load index, electromyographic activity, etc.) to a previously determined FH. In this sense, we can have scenarios with different intensity criteria for the same FH. E.g., given a DJ50, if we compare an athlete (a) who displays a concentric GRF of 4298 N, 3.15 of RSI, and a power output of 6452 W, to another (b) who has 3954 N of concentric GRF, 2.54 of RSI and 6462 W of power, it is clear how they may require different FH for maximizing concentric force production and reactive strength, but the same FH for maximum power output.

FH and its possible effects have been extensively addressed in the literature (Komi and Bosco, 1978a; Bobbert et al., 1987b; Matic et al., 2015a), but with still somewhat inconsistent results. The literature reports that the range of FH can be between 0.12 and 0.80 (Lees and Fahmi, 1994a; Viitasalo et al., 1998), although we know that higher FH have been studied (Clutch et al., 1983). Still, this can be considered a too wide FH range to be integrated in the training program of an athlete, given a training effect goal in a period of time. Therefore, using such a wide range of FH in a plyometric training program can be considered a methodological error. For plyometric training, we recommend that coaches consider a range of optimal FH, which should be related to the plyometric goal to be achieved, as shown in Figure 2.

In a training program, the FH contributes to training specificity (Ebben and Petushek, 2010) and enhances the magnitude of the different adaptations (Young, 1995). FH in excess of the individual capacity led to prolonged eccentric loading, which is synonymous with excessive plyometric loading. This causes longer GCT, which possible partially dissipate the accumulated elastic energy, thus preventing maximum jump height (Asmussen and Bonde-Petersen, 1974). Another element to consider when selecting the fall height is the principle of specificity. According to this principle, the way the athlete performs the plyometric activity will determine the training effect (Walshe et al., 1998). Furthermore, the higher the FH, the higher - up to a certain limit - the levels of neuromuscular pre-activation (Taube et al., 2012; Di Giminiani et al., 2020), the higher the speed that can be achieved in the eccentric phase, and thus the higher the contractile potentiating mechanisms of the muscle spindles and the elastic energy contribution by the tendons (Flanagan and Comyns, 2008). These mechanisms also depend on the individual's strength level, associated with maintaining active stiffness during the movement. All these conditions created by the increase or decrease in fall height are what can subsequently influence the intensity factor. Likewise, two athletes dropping from the same fall height may not be training at the same relative intensity (Byrne et al., 2017).

Although the literature recognizes the criteria discussed above, in this review, 90.99% of the included studies (Table 5) present arbitrary criteria for selecting the FH. Their procedures can be defined "arbitrary", as there are no *a priori* assessment criteria to determine which heights are to be used according to the individual characteristics of the subjects. In addition, there is no clear information regarding the relationship between the selected heights and the intervention goals for each subject. Some authors have recommended to limit plyometric exercises to heights of less than 40 cm, since they found that at heights between 40 and 60 cm, there was a reduction in propulsive power, vertical impulse and reactive jump height (Bobbert, 1990; Lees and Fahmi, 1994b; Voigt et al., 1995; Peng, 2011). However, this recommendation is highly questionable, as we consider that it cannot be generalized to comprise all types of population, including, for instance, high-level power athletes, who routinely use higher fall heights in their training programs (Kobal et al., 2017).

General discussion on plyometric intensity

Intensity is one of the most complex concepts within the theory and methodology of plyometric training. There is still no definitive consensus among researchers about the determination of plyometric training intensity, making it harder to be defended. In order to understand the multidimensional value of plyometric intensity, we need to revise the results of the most current studies on the topic.

It has been stated that plyometric intensity depends on the amount of tension exerted on muscles, connective tissue and joints (Kossow and Ebben, 2018b). Based on this statement, a DJ from 70 cm may not be considered high intensity, if the muscle actions are not performed to the maximum, or, on the contrary, it may be considered excessive intensity, if they are inhibited due to too high eccentric loading (Ruan and Li, 2010). Hence, intensity should be directly associated with a specific metric and not only with the FH. On this topic, research has suggested that intensity can be examined

based on different kinetic parameters (Ebben et al., 2008), others made contributions by analyzing the relationship between fall height and intensity (Bobbert et al., 1987b; McNitt-Gray, 1993; Young et al., 1999; Walsh et al., 2004; Wallace et al., 2010). For example, one study (Bobbert et al., 1987b) found no significant biomechanical differences in the joints during the eccentric phase of a DJ20 and a DJ40, but the eccentric phase maximum power was lower in the DJ60 than in the DJ40, and the DJ60 and DJ20 had similar values. Another study (Wallace et al., 2010) has found that eccentric GRF increases are directly proportional to the FH. This finding is likely due to gravity's uniform acceleration and the landing strategies of the subjects. If the athletes consciously hit the platform at landing, the eccentric forces will increase considerably. Our recommendation for high-performance athletes is to evaluate the GRF of the concentric phase, which can hardly be influenced by other variables other than the purely concentric forces of the athletes. Considering these parameters, we cannot disregard the power output as a significant plyometric intensity value (Di Giminiani and Petricola, 2016; Morin et al., 2019).

On the other hand, although the EMG is criticized for its lack of practicality in applied settings (Ramirez-Campillo et al., 2020a), some studies consider it the most important way of assessing plyometric intensity (Kossow and Ebben, 2018b). This is consistent with several studies that found higher EMG activity when performing a DJ60 and lower activity with a DJ20 (Peng et al., 2011), attributing a higher level of intensity for the DJ60. Plyometric exercises performed with a flexion at the knee joint, show higher EMG activity compared to rebounding exercises with no flexion at the knee joint (Jarvis et al., 2016). The same group of researchers evaluated the vastus lateralis and biceps femoris muscle activity during the concentric phase of the CMJ and the DJ. No significant differences were found for these muscle groups during both exercises. This could indicate that these two exercises, for what concern fibers recruitment during the concentric action, do not present significant statistical differences in intensity, yet it does not mean that they produce the same jump performance, due to the difference in elastic energy storage (Ishikawa et al., 2006). These results are consistent with other criteria previously investigated (Wallace et al., 2010; Sugisaki et al., 2013). With a different view, studies have been criticized for classifying plyometric exercises into low, medium and high intensity (Fowler and Lees, 1998; Potach and Chu, 2008). Although this classification (based on a gradual increase in intensity level) seems to make logical sense, research showed different results in lower limb mechanical performance depending on the type of plyometric exercise (Sugisaki et al., 2013). This is also reflected in other studies examining the EMG (Kossow and Ebben, 2018b). Previous research has showed that EMG activity for the Tuck Jump is highest for the muscles around the ankle. At the same time, only moderate levels of the neural drive are present for the muscles around the knee. These results show that the EMG activity should be considered when we want to know which exercises generate a lower or higher level of activation, which is undoubtedly a criterion for plyometric intensity.

It has also been recommended (Fowler and Lees, 1998) that GRF and momentum be considered factors influencing plyometric intensity, as it allows comparison of different exercise variants. Furthermore, it has been reported that there are no differences in these indicators at the neural level when unilateral and bilateral exercises are performed (Jarvis et al., 2016). However, some level of detail may be lost if only impulse is used to measure intensity, because the maximum level of mechanical stress experienced by the athlete may be consciously hidden (Jarvis et al., 2016). Furthermore, other research corroborates differences between the momentum of different plyometric exercises (Ebben et al., 2010). Therefore, it is recommended to perform a combination between the GRF and the EMG as evidenced in different studies (Van Lieshout et al., 2014; Kossow and Ebben, 2018b). Another important element that should be considered during DJ for reactive purposes is that the mechanical impulse is equal to the product of force times. Thus, the amount of force may be high, but if the amount of time is considerably increased, the reactive strength may be affected. Therefore, the impulse may not be a recommendable metric for analyzing reactive jumps. These means of intensity evaluation are helpful and are constantly evolving, yet they are not part of the practical reality of the sports coach domain. The dynamics of the implementation of plyometric programs do not allow these evaluations to become part of a daily training routine. The reduced rest intervals between exercises, large workgroups, and the unmet need for full-time specialized professionals during training, limit the presence and rigor of valid protocols.

In line with the previous statement, some researchers have proposed to assess training intensity based on contact time (Walsh et al., 2004; Ball et al., 2010), but jump technique (Bobbert et al., 1987a) and individual strategies to minimize GCT may mask the overall intensity values. On the other hand, the height achieved in the jump also has been suggested to assess intensity (Matic et al., 2015b), but this is also influenced by technical and individual strategies (Bobbert et al., 1987a; Bobbert et al., 1987b). Ideally, athletes should maximize JH while minimizing ground contact time. The RSI (Young, 1995) is the proposed metric that solves these problems through the use of different devices that can be easily apply during training (Montoro-Bombú et al., 2022). So far, we have not found in the literature any study assuming that the RSI can be considered a plyometric intensity variable. However, in its assessment, all the characteristics needed for its calculation are present for intensity evaluation. RSI is usually calculated by dividing the jump height by the ground contact time of a DJ (Ebben and Petushek, 2010). Among the studies that evaluated RSI, some implemented a training program (Ramirez-Campillo et al., 2016; Ramirez-Campillo et al., 2018b) and other used a double-blind controlled trial (Bogdanis et al., 2019; Jeffreys et al., 2019; Falch et al., 2022). These studies based their evaluations on traditional jumping tests (DJ and CMJ) and fall heights of ≤40 cm. In one of the studies (Ramirez-Campillo et al., 2019), athletes were required to achieve a contact time of ≤250 m, and failure to do so resulted in retesting. However, the RSI is generally considered highly modifiable, as decreasing contact time compromises jump height as well as the opposite (Newton and Dugan, 2002; Healy et al., 2018). Therefore, to maximize the RSI, the athlete must express an optimal relationship between ground contact time and reactive jump height (Healy et al., 2018). To establish a correct RSI value, it is recommended that researchers and practitioners consider these criteria in future research and carefully familiarize participants with this method prior to the assessment, in order to achieve valid and reliable plyometric intensity scores. Another point to consider is the standardization of the RSI equations. In fact, the

values resulting from dividing the JH by the GCT (Young, 1995) are not the same as those resulting from dividing FT by the GCT (Healy et al., 2018); the latter is always higher.

Different research has assessed plyometric intensity using the RPE (Asadi, 2014; Khodaei et al., 2017), and these authors recommend the RPE as a feasible method to assess plyometric intensity. Although using the earlier suggested methods to measure plyometric intensity is more popular among scientists, the necessary instruments are not available to strength and conditioning professionals (Asadi, 2014). For this reason, the research suggests that the RPE provides an innovative and practically compatible alternative for measuring plyometric exercise intensity. One investigation (Asadi, 2014) found a higher RPE for the SJ compared to the DJ35. These results contradict another study showing that the SJ had a lower perceived intensity score than the DJ, while the mean perceived intensity scores of Tuck Jump, Box Jump, the SJ and the DJ were just "moderate" on the RPE scale (Khodaei et al., 2017).

It should be noted that these studies on the RPE take a stand on plyometric intensity without any means of comparison (EMG, RSI, or force platforms). This is of concern, given that previous research (Müller et al., 2020) has found a weak relationship between RPE and reactive jumps power. Therefore, there is a need for further studies assessing the reliability and validity of the RPE for determining plyometric intensity. This is the reason why we did not include the RPE as a method for measuring plyometric intensity in this review. All these criteria help us reinforce the view that plyometric training intensity is a multidimensional factor. Human performance is so diverse that focusing the intensity measurement on a single variable is unwise. It is up to the coach or sport scientist to decide which parameters they want to measure within the training program, and at which stage of the training process it is more convenient to evaluate some parameters rather than others. For example, the DJ followed by a horizontal reactive jump has been shown to have better transfer to the acceleration over short distances, compared to the DJ followed by a vertical reactive jump which has better transfer over longer sprint distances (Meylan et al., 2009; Krejac et al., 2020). Furthermore, there may be no relationship between maximal eccentric GRF production and RSI, because eccentric GRF is favored by increased FH (Peng et al., 2017), while the highest RSI value requires an optimal FH. These two metrics can be assessed for different purposes and at different stages of the training process. Although we do not dismiss the possibility of knowing the maximum GRF during the best RSI performance, these values represent two different criteria for the evaluation of plyometric intensity.

Considerations on training volume in the included studies

Volume is a component of the training load that must be correctly established. Previous authors conceptualize volume as the total amount of activity to be performed, and it can be constituted by the sum of the work in sessions, weeks, months and years (Bompa and Buzzichelli, 2019). In the case of plyometric training, volume is also a primary criterion. It must be correctly related to the objectives to be achieved, stage of the training process, individual characteristics of the athletes and the metrics related to intensity.

Regarding training volume, a meta-analysis (de Villarreal et al., 2009) recommends a duration of 10 weeks for plyometric training programs and reports that interventions with more than 20 work sessions and more than 50 jumps per session delivered the best results. In the same line of research (Sáez de Villarreal et al., 2012) also indicated that excellent results could be observed in less than 10 weeks (6–8 weeks), reporting that 18 work sessions and more than 80 jumps per session seem to deliver positive adaptations. Even though multiple researchers (Gehri et al., 1998; Luebbers et al., 2003; Markovic et al., 2007; de Villarreal et al., 2008; Chelly et al., 2010; Ramírez-Campillo et al., 2013; Chelly et al., 2014; Ramírez-Campillo et al., 2014; Terzis et al., 2016; Ramirez-Campillo et al., 2018d; Bianchi et al., 2019; Bogdanis et al., 2019; Jeffreys et al., 2019; di Cagno et al., 2020; Ahmadi et al., 2021; do Carmo et al., 2021; Sanchez-Sixto et al., 2021; Falch et al., 2022; Lum et al., 2022) share these positions, we consider training volume an eminently individual parameter. The previously mentioned studies (Table 5) only report group averages in their results, but there is the need to conduct more studies analyzing the individual volume response. The principle of training individualization (Bompa and Buzzichelli, 2019) supports this position. Predetermined volumes may favor some athletes by providing them with an adequate stimulus, but may also be detrimental to others by giving an insufficient or excessive stimulus. We should not overlook the fact that the above mentioned studies based their recommendations on the basis of different measured metrics. In this sense, there might be a difference in training volume to improve the GRF *versus* to improve the RSI. However, this criterion requires further investigation.

Another characteristic commonly observed in the intervention programs was a linear increase of the workload over time. This is widely suggested in the theory of sports training (Bompa and Buzzichelli, 2019), and also responds to the principle of progressive overload (Kasper, 2019), which, in the case of plyometric training means an increase in the tension exerted (Ramirez-Campillo et al., 2020c). Nevertheless, it could be considered a methodological error to establish weekly (Ahmadi et al., 2021) or fortnightly increments (Ramirez-Campillo et al., 2018d; Ahmadi et al., 2021; do Carmo et al., 2021) without prior assessments that would guarantee the sought after adaptations at individualized heights and work volumes. None of the plyometric programs included justifies the rationale for the timing and magnitude of the increase of the load. This review assumes that the gradual increase of the load responds to a criterion of adaptation and control, thus we only recommend increasing the load when each individual athlete shows a positive adaptive response. All the studies included in this systematic review (Table 5) selected jumps volume in an arbitrary and standardized way for all participants. This is undoubtedly a point that needs to be reconsidered in all future research.

All the criteria previously discussed, together with suggestions and conclusions of previous studies (Gehri et al., 1998; Luebbers et al., 2003; Markovic et al., 2007; de Villarreal et al., 2008; de Villarreal et al., 2009; Sedano et al., 2009; Chelly et al., 2010; Khlifa et al., 2010; Sáez-Sáez de Villarreal et al., 2010; Ramírez-Campillo et al., 2013; Chelly et al., 2014; Ramírez-Campillo et al., 2014; Terzis et al., 2016; Eagles et al., 2018; Ramirez-Campillo et al., 2018d; Bianchi et al., 2019; Bogdanis et al., 2019; Jeffreys et al., 2019; Silva et al., 2019; di Cagno et al., 2020; Laurent et al., 2020; Ramirez-Campillo et al., 2020c; Ahmadi et al., 2021; do Carmo et al., 2021;

Ramirez-Campillo et al., 2021; Sanchez-Sixto et al., 2021; Falch et al., 2022; Lum et al., 2022), allow us to list the following methodological recommendations that may be useful for the organization of new plyometric training programs for coaches and researchers.

Conclusion

Several problems regarding the methodological organization and procedure of the proposed plyometric training programs were identified, such as the lack of consensus on a training intensity criterion, a lack of justification for the distribution and orientation of the training volume, as well as a lack of criteria for the individualization of the training process. For these reasons, we established criteria which we consider fundamental for prescribing plyometric training programs. A new criterion for the identification of the subjects' experience level was established, as well as methodological recommendations for the selection of training intensity and training volume. It is assumed that the drop height should not be understood as the sole determinant of the intensity of plyometric training but should be an integrated factor to determine intensity. These results could be helpful for sport scientists and sport coaches who intend to improve and implement new plyometric training programs. It is also recommended that researchers considered the above recommendations when conduct research procedures.

Scientific-methodological considerations for the implementation of new plyometric and ballistic training programs

1. Soft surfaces, being more associated with the production of maximal dynamic strength, can be used for the drop jump, when a greater emphasis on countermovement and jump height is sought and when short contact times are not required (DJs with countermovement or Depth Jumps).
2. Soft surfaces can be used for rehabilitation, during the plyometric preparation phase or general preparation phase of athletes, as the training in those phases does not require short ground contact times.
3. Soft surfaces can be more adequate during the first developmental phase of sprint acceleration, where more work on high-force production at relatively low speeds is sought, as well as for quintuple and deca-jumps in general preparation. In addition, they can be used with a wide variety of jumps which constitute the general preparation of top athletes.
4. During training on soft surfaces, athletes should be guided to eliminate heel strikes against the ground so that these are not transferred to hard surface work.
5. Hard surfaces are recommended to be introduced after working on soft surfaces.
6. Hard surfaces are associated with short ground contact times, and bouncing jumps with open knee joint angles.
7. It is recommended not to instruct the landing as "straight leg". Substitute that instruction with "bouncing fast off the ground with the smallest possible knee angle" or a similar orientation.
8. Hard surfaces are more associated with the development of power, muscle stiffness and reactive strength, so they should be introduced gradually in specific preparation, and kept being used in the pre-competitive and competitive phases.
9. A wide variety of jumps that require short contact times and are primarily based on fast bouncing, can be introduced while training on hard surfaces, as long as metrics such as maximum power output, best reactive strength, etc., are being monitored.
10. Plyometric exercises with less rapid contact times, involving a countermovement and higher drop heights, aiming at producing maximal dynamic strength, increases in ground reaction forces or maximal eccentric force production, can be combined with general or maximal strength work. In this case, Depth Landings, Depth Jumps, Horizontal Drop Jumps, and jumping between high hurdles are recommended.
11. Plyometric exercises with more open knee and hip angles, fast contact times and optimal drop heights, used for the development of maximum power output production, RFD, and Reactive Strength, can be combined with power-oriented weight training work. In this case, exercises such as Tuck Jumps, Drop Jumps, and jumping between medium hurdles are recommended.
12. The combination of fast and slow plyometric exercises (rebounding and countermovement) can be introduced gradually in the stages of transition from general to special preparation or have a specific application during the preparation. For example, if plyometric jumps are used to reinforce maximum dynamic strength during special preparation, or the competitive or tapering phases, Drop Jumps can be combined with horizontal Drop Jumps or Depth Jumps.
13. To determine the level of experience of the participants in a plyometric training program, we recommend the use of the following equation:

$$\text{Level of plyometric experience} = [(A + B + C/2)/3] \times (D + E)$$

14. The training intensity can be individualized by determining (evaluating) the optimal height for the highest power output, reactive strength, RFD, ground reaction forces, stiffness, reactive jump height, or other performance parameters.
15. Training programs for large groups of athletes should contain different drop heights so that each athlete can train according to his or her level of adaptation. In addition, this facilitates work organization as the group is distributed over several plyo boxes.
16. Training volume can be individualized using two strategies: a) monitoring the individual session so that each athlete does not increase ground contact times, or lose power output or reactive strength, while seeking to maintain the best values for a given metric. Alternatively, b) by establishing a percentage of performance loss compared to the best result in each metric. This ensures that the athlete can train with different performance orientations (as required by the coach), at the same time, it allows athletes to work according to their daily capabilities without having a pre-

established number of sets or repetitions that can sometimes be excessive or inadequate.

17. The selection of the optimal drop height should be determined via a specific performance metric. For instance, plyometric training programs aiming at the improvement of maximum power production should establish the DJ fall height that maximize that metric.
18. For a correct execution of the drop, it is recommended that the athlete stands at the edge of the box (not in the center), determines a starting leg, directs the toe up and lets the body fall freely (without pushing forward).
19. We recommend that both the drop height and the training volume (sets and repetitions) should only be increased when real adaptations to the proposed training variables are observed.

Author contributions

R-BM and CB were involved in conceptualization, data Collection Process and sources of information. R-BM and HS in the eligibility criteria, search strategy methodology. AS participated in the formal analysis. LR and JG were involved in the supervision. LR and CB translation of the paper. R-BM and LR participated in the writing of the original draft. All authors were involved in reviewing and editing the draft. All authors read and approved the final version.

Conflict of interest

Author NM was employed by the company Chinese Athletics Association.

The remaining authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Supplementary material

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fphys.2023.1181781/full#supplementary-material

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Lower limbs inter-joint coordination and variability during typical Tai Chi movement in older female adults

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Purpose: This study aimed to investigate the lower limb inter-joint coordination and variability during Tai Chi movements compared with normal walking in older adults.

Methods: A total of 30 female Tai Chi practitioners (70.9 ± 5.2 years) were recruited in this study. Herein, each participant performed three trials of the normal walking and Tai Chi movements. The lower limb kinematics data were collected with Vicon 3D motion capture system. The continuous relative phase (CRP) includes both spatial and temporal information of two adjacent joints, which was calculated to assess the inter-joint coordination of lower limbs. Coordination amplitude and coordination variability were assessed with mean absolute relative phase (MARP) and deviation phase (DP). MANOVOA was used to analyze inter-joint coordination parameters between different movements.

Results: The CRP values of hip-knee and knee-ankle segments in the sagittal plane of the Tai Chi movements changed frequently. The MARP values of the hip-knee ($p < 0.001$) and knee-ankle segments ($p = 0.032$) as well as the DP values of the hip-knee segment ($p < 0.001$) were significantly lower in Tai Chi than in normal walking.

Conclusion: More consistent and stable inter-joint coordination patterns of Tai Chi movements found in this study may be one of the critical factors that Tai Chi could be a suitable coordinated exercise for older adults.

KEYWORDS

older adults, Tai Chi, inter-joint coordination, continuous relative phase, phase plot

1 Introduction

Falls are a major health problem for older adults worldwide. The incidence and number of falls in elderly women were about twice as high as those in males in the United States (Blazewick et al., 2018). Falls can result in severe injuries, particularly in postmenopausal female adults with a risk of osteoporosis, such as soft tissue injuries, craniocerebral injuries, as well as fractures, and death among older adults (Mateen and Király, 2016). Losing balance is one of the most frequent cause of falls in the elderly (Fabio et al., 2004). Falls prevention requires precise posture control stability, which can be highly related to inter-joints coordination patterns while maintaining body balance (Hsu et al., 2014). However, inappropriate control of the joints of the locomotor system may contribute to body imbalance which may place an individual at risk of falls (Hsu et al., 2016). Therefore,

interventions are needed to help the elderly develop strategies for improved inter-joint coordination, particularly female adults.

Thus, Tai Chi is a promising method. As a low-speed, low-impact exercise, Tai Chi is a traditional Chinese exercise that is popular among older adults. In addition, Tai Chi is beneficial for retaining or regaining proper balance and coordination for older people (Kuo et al., 2021). A six-week Tai Chi intervention can improve functional movement coordination, such as placing, turning, and displacing (Wu et al., 2004; Burschka et al., 2014). Given the variable movement velocity between two joints or segments, the kinematic and kinetic analyses of single joints may be insufficient to reveal the coordination characteristics of limbs (Chiu et al., 2010) Inter-joint coordination, which is used to assess the relationship between angular positions and the velocities of two joints, can capture the underlying multiple-joint coordination dynamics in motor tasks to overcome insufficient single-joint coordination (Chiu and Chou, 2012). Therefore, understanding the inter-joint coordination of multiple joints also provides insightful information about postural control and fall prevention in human motion (Mehdizadeh et al., 2015).

Continuous relative phase (CRP) is a commonly used technique to investigate movement coordination. CRP is derived from the angular velocity–angle phase diagram of two joints or segments changing with time, and it is associated with information about afferent joint receptors (Chiu and Chou, 2012). The magnitude and variability of CRP curves are assessed by calculating the mean absolute relative phase (MARP) and deviation phase (DP). In crossing obstacles, Lu et al. observed that during the swing phase, the DP values of the leading limb were smaller than those of the trailing limb; however, the coordination of the leading limb was more stable than that of the trailing limb (Lu et al., 2008). Moreover, limbs with low DPs value show adaptability to environmental changes and have a stable coordination pattern during obstacle crossing of the elderly, concussed, and bilateral osteoarthritis groups (Chiu et al., 2013).

However, the multi-joint coordination of lower limbs is lacking during typical Tai Chi movements, thus more scientific research is needed to understand the basic mechanisms and principles of how Tai Chi works. Therefore, this investigation aimed to describe and quantify the lower limb inter-joint coordination and coordination variability during Tai Chi movements in older adults and explain the beneficial effect of Tai Chi on coordination and proprioception compared with normal walking. For comparison, the normal walking gait is chosen because it is the most common gait pattern in daily activities in elderly populations. In addition, the brush knee and twist step (BKTS) is a typical basic Yang-style Tai Chi movement and continuous forward gait, with gait phases similar to normal walking (Li et al., 2019). It was hypothesized that (1) more tightly coordinated inter-joint coordination dynamics occur during Tai Chi movements than during walking; (2) Tai Chi movements have less variable inter-joint coordination.

2 Materials and methods

2.1 Participants

A total of 30 healthy female Tai Chi practitioners (age: 70.9 ± 5.2 years, height: 161.4 ± 6.1 cm, body weight: 62.6 ± 8.7 kg, and Tai Chi experience 12.0 ± 5.1 years) were recruited in this study. Inclusion criteria are as follows (Li and Law, 2018): (1) aged 65 years or over; (2) no lower limb injuries; (3) can walk independently and complete the test; and (4) exercise duration of more than 5 years (Song et al., 2017; dan et al., 2018). Meanwhile, exclusion criteria are as follows: inability to follow instructions, heart conditions, joint replacements in the lower extremities, arthritis, visual impairments, vestibular disorders, or any neuromuscular problems preventing the participants from meeting the project requirements. All participants signed an informed consent agreeing to participate. This study has been approved by the Ethics Committee of Shandong Sport University (No. 2017103).

With a sample size of 30, the power of all statistical comparisons in this study was no less than 0.80. It was reported that 15–20 participants were sufficient in estimating the reliability of a quantitative variable (Fleiss, 1986).

2.2 Apparatus

A walkway was constructed for data collection (Figure 1). Two force plate (Kistler, 9287BA, Winterthur, Switzerland) were embedded into the walkway, and the ground reaction force was collected at 1,000 Hz (Song et al., 2017). The height of the force plate is consistent with the walking path. The kinematics data of lower limbs were collected with an 8-camera Vicon motion capture system (Vicon, Inc., Oxford, UK) at a sampling frequency of 100 Hz (Song et al., 2017). The force and kinematic data collection were synchronized using the Vicon system.

2.3 Data collection

The participants' anthropometric data were collected before the tests. Herein, an experienced research lab technician and participants were blinded to conduct data collection. Then, each participant was asked to wear a tight black uniform and shoes provided by the laboratory. A total of 41 infrared reflective markers with a diameter of 14 mm were attached to the skin or clothes of each participant with double-adhesive tapes. Afterward, before data collection, the participants were instructed to perform 5 min warm-up to familiarize themselves with the experimental set.

The foot movements during the BKTS are described as follows: forward movement, the right foot makes contact with the ground first, the left foot then steps forward, and the right foot leaves the ground at the end (Figure 2). Then, the participants were instructed to perform randomly-assigned tasks independently at a self-selected pace during the experiment. Each subject was asked to practice three times and adjust the starting point of each gait so that the subject's dominant leg could consistently and naturally contact one force plate at the beginning and another force plate at the end of a complete gait cycle. Once ready, each subject performed three successful trials of each movement (Mao et al., 2006). A successful trial was considered if the subject stepped correctly on all the force plates with a natural and smooth movement, complete gait cycle, and no loss of markers. This was determined by the investigators who observed the subject's performance. If a trial was not valid, it was discarded in data collection and not used for the subsequent statistical analyses.

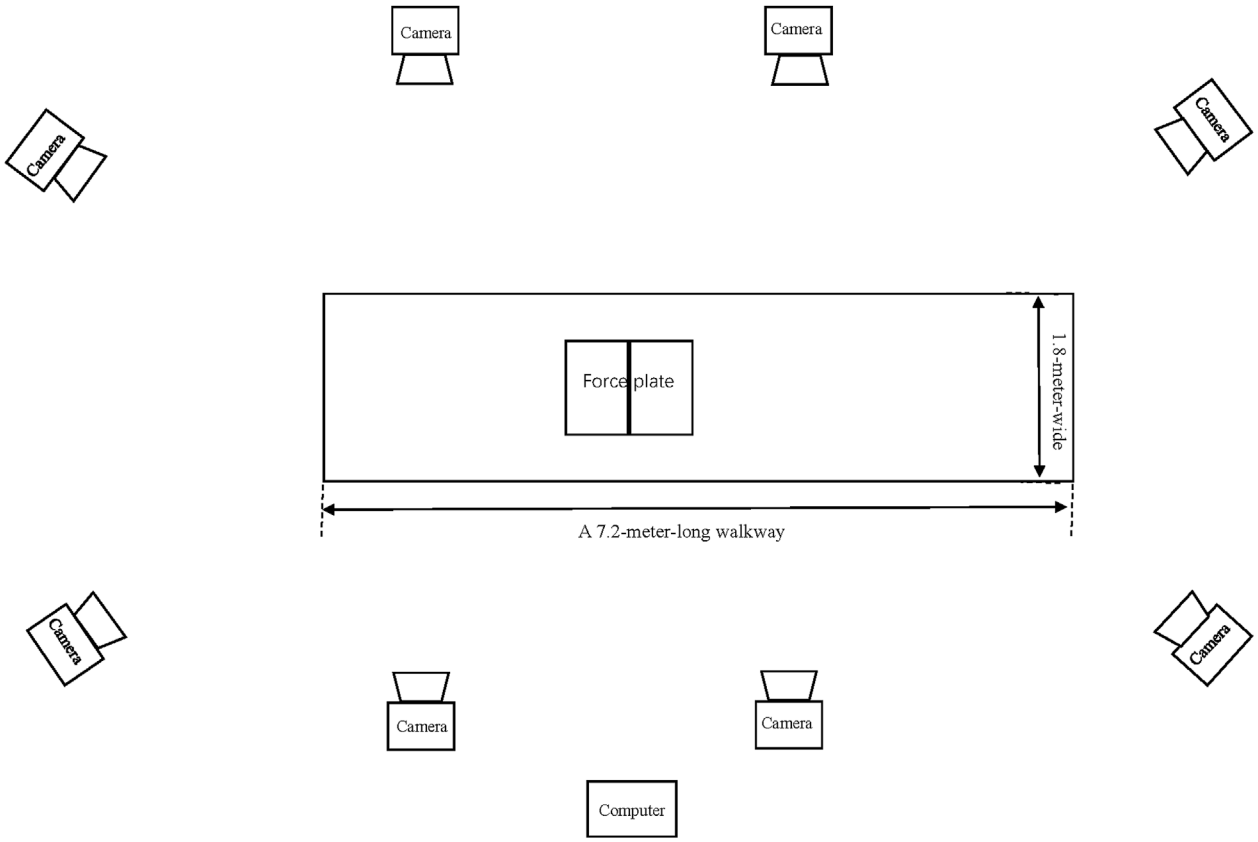

FIGURE 1
Demonstrations of experimental site.

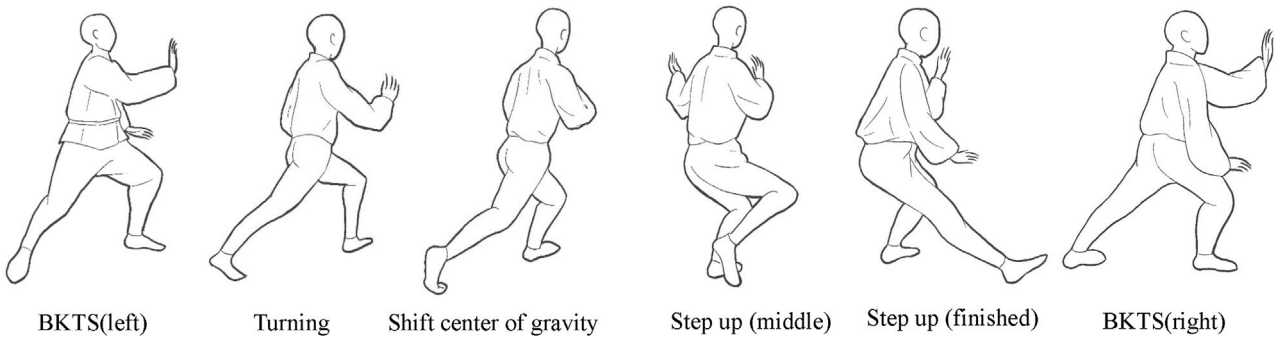

FIGURE 2
Graph for brush knee and twist step.

2.4 Data processing

As determined through a kicking ball test, the right leg was the dominant limb among all participants. In addition, the left and right cycles have quasi-identity during most Tai Chi movements (Mao et al., 2006). Therefore, only the data from the dominant leg was selected for analysis. Besides, given that the kinematic changes of lower extremity joint associated with walking and age-related primarily occur in the sagittal plane (Huang et al., 2020), we only examined the sagittal plane inter-joint coordination between the hip and knee and between the knee and ankle. The gait cycle was defined from the right foot landing on the ground to the same foot landing again for the Tai Chi movement and normal walking. Then, the kinematic parameters of one gait cycle for lower limbs were analyzed, and each gait cycle was normalized to 100 equal time intervals.

2.4.1 Phase plot

The phase portrait for each joint throughout a gait cycle was generated by plotting the normalized angular positions (θ) (*x*-axis) and velocities (ω) (*y*-axis). Angular positions and velocities were normalized using the following equations:

$$\theta_i = \frac{2 \times [\theta_i - min\ (\theta_i)]}{max\,(\theta_i) - min\,(\theta_i)} - 1$$

$$\omega_i = \frac{\omega_i}{max\,|\omega_i|}$$

where θ_i and ω_i are the angular position and velocity for each data point during a gait cycle, respectively. Such normalization defined the angular positions and velocities between 1 and -1 along both dimensions of the phase plane and minimized individual differences in amplitude and frequency (Tung-Wu et al., 2008). Phase angles were calculated using the following equation at each time point:

$$\varphi = tan^{-1}\left(\frac{\omega}{\theta}\right)$$

2.4.2 Continuous relative phase (CRP)

The CRP, which identifies the coordination between two adjacent joints, was obtained by subtracting the phase angle of a distal joint from that of a proximal joint (φhip-knee, φknee-ankle) (Lamb and Stöckl, 2014). A positive value indicates that the proximal joint leads to the distal, and *vice versa* (Lamb and Stöckl, 2014). If the Relative Phase Angle (RPA) is close to 2π (0°or ±360°), the two joints are moving in a similar fashion or phase. Meanwhile, the RPA is close to π (180°), the two joints are moving in an opposite fashion or out of phase.

2.4.3 Coordination variability

Mean absolute relative phase (MARP) was calculated by averaging the absolute values of the ensemble curve points for the tasks (Stergiou et al., 2001):

$$MARP = \frac{\sum_{i=1}^{P} |\varphi|}{P}$$

where P is the number of points in the periods. Functionally, a low MARP value indicates a more in-phase relationship between the action of two joints or segments for a task and a given subject (Ippersiel et al., 2021).

The deviation phase (DP) was calculated by averaging the standard deviations of the ensemble RPA curve points for the tasks (Stergiou et al., 2001; Hein et al., 2012):

$$DP = \frac{\sum_{i=1}^{P} SD}{P}$$

Functionally, a low DP value indicates a stable relationship between the action of two joints or segments for a task and a given subject (Ippersiel et al., 2021).

2.5 Statistical analysis

The average values of three successful trials were used for statistical analysis. The multivariate analysis of variance (MANOVA) was used to compare the inter-joint coordination parameters between Tai Chi movements or normal walking. The significance level was set at 0.05. All the statistical analyses were performed with SPSS software 20.0 (IBMS, NY, USA).

3 Results

The flexion and extension angles for the joints changed more frequently in Tai Chi movements as compared with normal walking (Figure 3). As shown in Figure 4, the ensemble-averaged phase plots of the lower limb joints during normal walking and Tai Chi. During normal walking, the hip, knee, and ankle trajectories were all in form of nearly closed and period circles, whereas those for the Tai Chi movements had a more complex form of trajectories with varying amplitudes.

Figure 5 shows the ensemble-averaged hip-knee (φhip-knee) and knee-ankle (φknee-ankle) CRP of the lower limb joints during normal walking and Tai Chi movements. However, the leading joint between the hip and knee segments and between the knee and ankle segments changed more frequently during Tai Chi as compared with normal walking. During Tai Chi movements, the hip-knee CRP curves showed a W-shape with two negative peaks during the end of gait cycles, whereas the knee-ankle CRP curves showed an M-shape with two positive peaks. Similar shapes were also found for the CRP curves of normal walking, except those of curve changes in Tai Chi movement which lagged behind those of normal walking and the smaller peaks of Tai Chi.

Table 1 provides the MARP and DP values of the CRP curves (hip-knee and knee-ankle) for normal walking and Tai Chi movements. During Tai Chi movements, the MARP and DP values of the hip-knee CRP curves were significantly smaller than those of normal walking ($p < 0.001$). During Tai Chi movement, the MARP values of the knee-ankle CRP were significantly smaller than those of normal walking ($p = 0.032$).

4 Discussion

This study was the first to examine the inter-joint coordination of older adults while performing Tai Chi movements. The present study results exhibited that the inter-joint coordination between the hip and knee and between the knee and ankle moved nearly in phase or synchronized during Tai Chi movement. Furthermore, low variability of hip-knee and knee-ankle coordination was observed during Tai Chi movements. These findings indicate that the Tai Chi movement has better and more stable inter-joint coordination patterns than normal walking. Therefore, the results of this study supported our hypothesis. The result also agrees with previous studies showing that older adults that regularly practice Tai Chi exhibit better postural stability than those who do not (Wu et al., 2004).

In addition, the lower-limb joint-angle changes were more frequent during Tai Chi movement as compared with normal walking. During Tai Chi movement, the present study also found that the phase plot of the three joints of the lower extremities appeared in extra closed circles. Based on previous studies, the phase plots in Figure 4 represent a stable, periodic motion given that velocity and position repeatedly return to a certain value (Lu

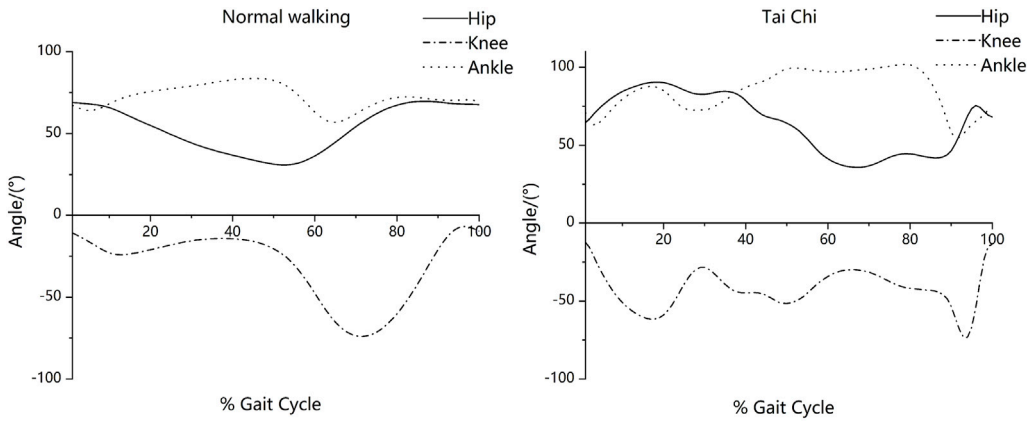

FIGURE 3
Angles of lower limb joints in the sagittal plane for normal walking and Tai Chi movements.

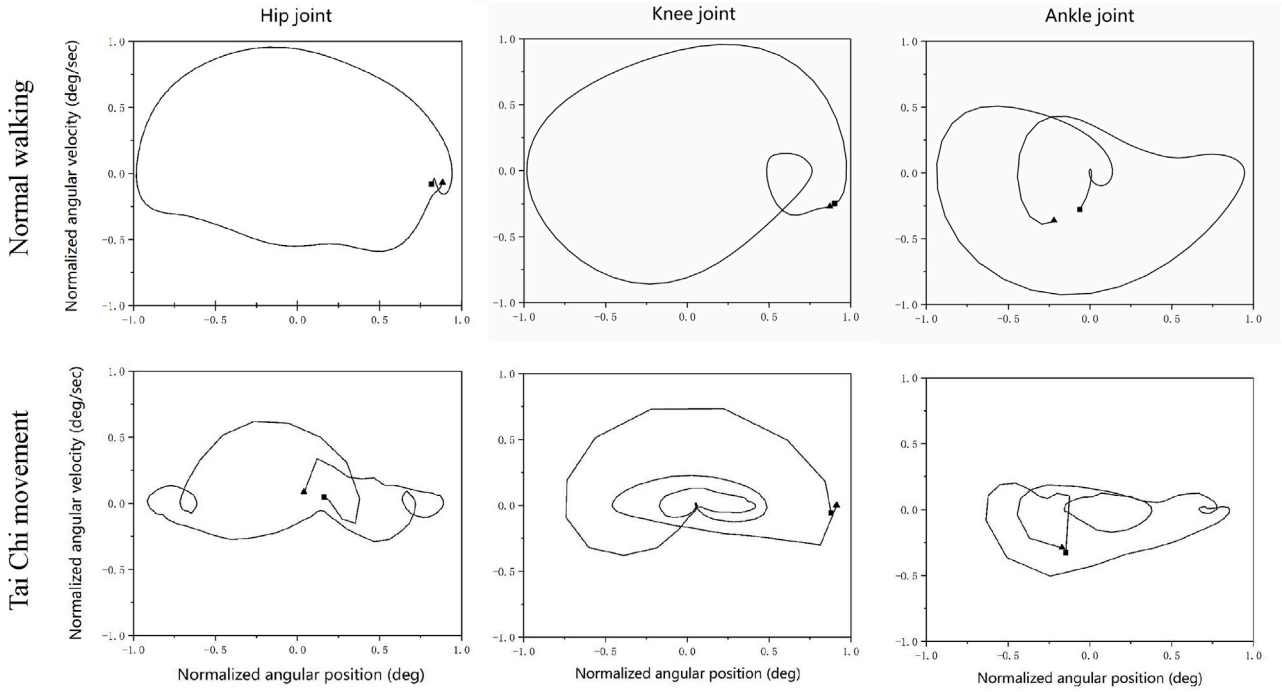

FIGURE 4
Ensemble-averaged phase plots of the hip, knee, and ankle of the lower limb in the sagittal plane for normal walking and Tai Chi movements. The longitudinal axis is normalized angular velocity and the horizontal axis is a normalized angle.

et al., 2008). Both Tai Chi movement and normal walking are continuous forward movements, but the first one is more complex. During Tai Chi movement, the first step forward, the center of gravity (COP) gradually transitioned to the front limb; then the back limb bent the knee and the front limb extended, the COP shifted to the back limb; and finally turned around, the back limb pushed off, the COP returned to the front limb again. Compared to normal walking, in Tai Chi movements the COP shifted from back to forward by changing the angle of the lower limbs. The motion characteristics may be closely related to the above results.

The frequent changes in joint angle indicate that the elderly need to adjust repeatedly, and accurately control the body's COP and limb position while performing BKTS. This may help regulate mechanoreceptors' input-output relationship and induce plasticity changes in the central nervous system (such as the increase of synaptic connection strength) (Ribeiro and Oliveira, 2007). This will also help strengthen signaling pathways and enhance

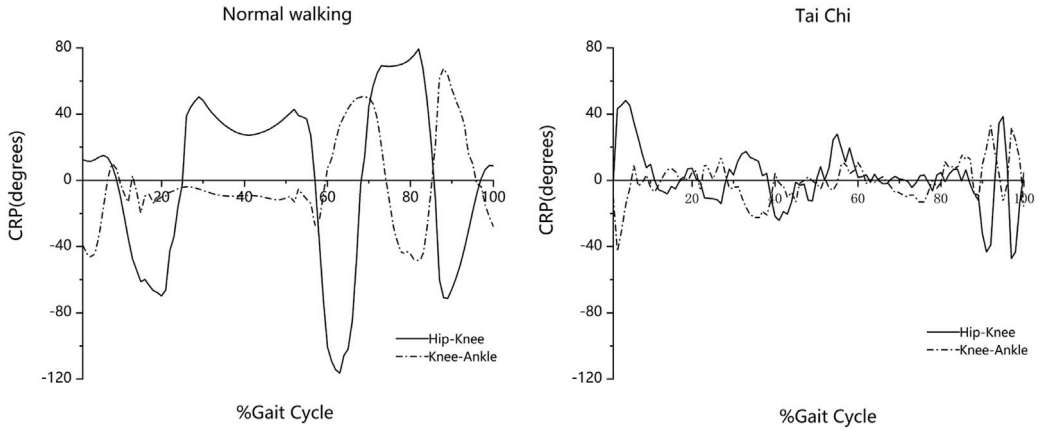

FIGURE 5
Continuous Relative Phase of the lower limb in the sagittal plane for normal walking and Tai Chi movements.

TABLE 1 Internal coupling relationship of the sagittal plane of the lower limb (Mean ± SD).

Value (°)	Normal walking	Tai Chi	*p* Value
Hip-Knee MARP	55.44 ± 3.84	36.18 ± 6.11	<0.001
Knee-Ankle MARP	35.54 ± 5.12	32.64 ± 5.09	0.032
Hip-Knee DP	0.64 ± 0.03	0.50 ± 0.06	<0.001
Knee-Ankle DP	0.49 ± 0.06	0.46 ± 0.05	0.069

proprioception (Ribeiro and Oliveira, 2007). Proprioception provides feedback on body position and movement that play important role in coordinating movement (Montel, 2019). Moreover, nearly all motile animals rely on proprioceptive feedback to control their bodies (Tuthill and Azim, 2018). The enhancement of proprioceptive conduction function provides more so-called "reflex" pathways for motor output (Tuthill and Azim, 2018), which may allow for more precise movement trajectory and more stable inter-joint coordination patterns of limbs, thereby reducing the risk of falls.

The patterns of the joint phase plot suggest stable inter-segment coordination for each movement. Therefore, the data on inter-joint coordination must be closely investigated for insights into the overall control of the lower extremity during Tai Chi movements. The dominant joint shifts more frequently during Tai Chi movements, thereby indicating changes in the proximal and distal joint-guided motion patterns. At the beginning of the movement cycle, the proximal joint flexes and acts as an active joint to guide motion. Subsequently, hip-knee and knee-ankle coordination gradually changed from out-phase to in-phase to maintain a COP stability. At the end of the movement cycle, the left leg kicks off, and the lower limb flexes to complete an action similar to crossing an obstacle, thereby making joint coordination and posture more difficult to control.

Proximal joints play an important role in postural control, and their dynamic adjustment may increase instability in the body (Rahimzadeh et al., 2020). However, MARP and DP values of hip and knee and knee and ankle CRP curves during the BKTS were lower than normal walking. The smaller the MARP values, the closer the segments are to the in-phase (Ippersiel et al., 2021). Moreover, the smaller the DP values, the lesser the variability, and the more stable the motion (Zhang et al., 2015). This indicates that the motion between the joints of the lower limbs is more synchronized and stable during BKTS, with less independent motion between the joints. This may indicate that older adults adopt more cautious gait strategies when rehearsing knee walking, thereby moving joints more coordinated by frequently adjusting the motion amplitude and speed of adjacent joints. Furthermore, joint coordination plays a crucial role in maintaining gait stability (Ippersiel et al., 2021). This frequent adjustment may be beneficial to optimize the proprioceptive information transmission of the human body and thus promote the formation of adaptive joint coordination patterns between hip-knee and knee-ankle joints. Therefore, BKTS may reflect a coordinated and stable gait control strategy, thereby maintaining posture stability during difficult motor tasks and may help reduce the risk of falls (Mehdizadeh et al., 2015).

However, this study encountered three limitations. First, we disregarded the effects on the trunk and upper limbs. Given that Tai Chi is performed with whole-body segment motions, thus these segments might have been affected. Second, the pattern and variability of inter-joint coordination were only examined in the sagittal plane. Given that the sagittal plane is the primary plane of motion during gait, the inter-joint coordination is expected to be robustly controlled and illustrated. Third, all speeds were self-selected. We intended to examine variations in the inter-joint coordination within a range of self-selected paces and minimize interferences in the participants' performance.

5 Conclusion

Tai Chi movement prompted better adaptability of movement controls and more tightly and stable inter-joint coordination patterns of lower limbs in older adults. This response may reflect

a cautious gait pattern, or a functional strategy to help mitigate the risk of falls in older adults. Therefore, Tai Chi exercise may provide a unique way of coordination training, particularly for the coordination between the segments, which may contribute to postural control and thus play a role in preventing falls.

Data availability statement

The raw data supporting the conclusions of this article will be made available by the authors, without undue reservation.

Ethics statement

The studies involving human participants were reviewed and approved by Ethics Committee of Shandong Sport University. The patients/participants provided their written informed consent to participate in this study.

Author contributions

JZ: provided research design, project management, data collection, data analysis and manuscript writing. WH: provided project management, data interpretation and manuscript modification. HT: provided project management, data collection, data analysis, data interpretation and manuscript modification.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Effects of EMG-based robot for upper extremity rehabilitation on post-stroke patients: a systematic review and meta-analysis

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Objective: A growing body of research shows the promise and efficacy of EMG-based robot interventions in improving the motor function in stroke survivors. However, it is still controversial whether the effect of EMG-based robot is more effective than conventional therapies. This study focused on the effects of EMG-based robot on upper limb motor control, spasticity and activity limitation in stroke survivors compared with conventional rehabilitation techniques.

Methods: We searched electronic databases for relevant randomized controlled trials. Outcomes included Fugl-Meyer assessment scale (FMA), Modified Ashworth Scale (MAS), and activity level.

Result: Thirteen studies with 330 subjects were included. The results showed that the outcomes post intervention was significantly improved in the EMG-based robot group. Results from subgroup analyses further revealed that the efficacy of the treatment was better in patients in the subacute stage, those who received a total treatment time of less than 1000 min, and those who received EMG-based robotic therapy combined with electrical stimulation (ES).

Conclusion: The effect of EMG-based robot is superior to conventional therapies in terms of improving upper extremity motor control, spasticity and activity limitation. Further research should explore optimal parameters of EMG-based robot therapy and its long-term effects on upper limb function in post-stroke patients.

Systematic Review Registration: https://www.crd.york.ac.uk/PROSPERO/; Identifier: 387070.

KEYWORDS

EMG-based robot, stroke, upper extremity, meta-analysis, review

1 Introduction

Stroke is a prevalent neurological dysfunction syndrome characterized by high incidence, mortality and disability rate (Winstein et al., 2016). Upper limb disorders are present in 85% of stroke survivors, with motor dysfunction still affecting 55%–75% of patients 3–6 months after onset (Parker et al., 1986; Feys et al., 1998). The residual upper limb dysfunction hugely impacts the ability of post-stroke patients to live and work independently, leading to reduced quality of life and a burden on patients' families and society (Micera et al., 2020). Therefore, there is an urgent

need to promote upper extremity function in post-stroke patients. However, the need for effective rehabilitation techniques for upper limb in stroke survivors remains largely unmet. Thus, it is of great significance to develop effective and positive rehabilitation methods for the upper limb rehabilitation of stroke survivors.

Conventional rehabilitation techniques, such as constraint-induced movement therapy (CIMT), physical therapy (PT) and occupational therapy (OT) (Pollock et al., 2014; Corbetta et al., 2015), have been adopted to assist upper limb rehabilitation. These techniques require patients to perform full or partial-assisted movements under the supervision of therapists. Other rehabilitation methods such as electrical stimulation (ES) and robots can provide repetitive, high-intensity training and also have benefit to reduce the physical stress of rehabilitation staff (Doucet et al., 2012; Zhang et al., 2017; Wang et al., 2021). However, a lack of real-time feedback from patients and excessive electrical stimulation may impede the efficacy of repetitive, high-intensity training and, in certain instances, induce muscle fatigue, thereby hindering the facilitation of motor function recovery (Chae et al., 2002).

Electromyography (EMG) have been utilized to control electrical stimulations (Hu et al., 2010; Rong et al., 2015; Rong et al., 2017; Nam et al., 2022) and powered exoskeletal devices (Rosen et al., 2001; Cheng et al., 2003; Dipietro et al., 2005; Ferris et al., 2005; Song et al., 2008) and trigger robot-assisted training to provide movement assistance (Zhuang et al., 2021). The EMG-based robot is one of the novel techniques designed for maximizing the involvement of voluntary efforts during post-stroke training. Unlike traditional robot-assisted training, the EMG-based robot can detect residual EMG signals of the affected limb in real time and integrate the participants' voluntary motor intention represented by the EMG signal from the residual muscles into training (Song et al., 2008; Rong et al., 2015; Chen et al., 2022). Once the EMG signals reach a specific threshold, the robot-assisted training will be activated, assisting the patients to complete a desired movement. The EMG-based robot could increase the interaction between participants and machines, potentially enhancing the effect of robot-assisted training, and reducing the pressure on medical staff. Studies have found that the EMG-based robot improved the Fugl-Meyer assessment (FMA) score and spasticity of the upper extremity in post-stroke patients when compared with conventional therapy, (Stein et al., 2007; Song et al., 2008; Hu et al., 2015; Nam et al., 2017). However, Chen et al., (Chen et al., 2022), and Page et al., (Stein et al., 2007; Page et al., 2013; Page et al., 2020), respectively found the efficacy of the EMG-based robot was not superior to task-oriented training and conventional hands-on manual therapy. Therefore, it remains unclear whether the effect of the EMG-based robot is superior to conventional therapies on upper limb function of stroke survivors.

To date, there is a lack of meta-analyses summarizing whether EMG-based robot training is superior to conventional treatment. Therefore, the objectives of this meta-analysis are twofold: 1) to determine the superiority of EMG-based robot therapy over conventional therapy, and 2) to analyze the effectiveness of different treatment options.

2 Methods

This review was on the basis of the Preferred Reporting Items for Systematic Reviews and Meta-Analyses (PRISMA) (Moher et al., 2009).

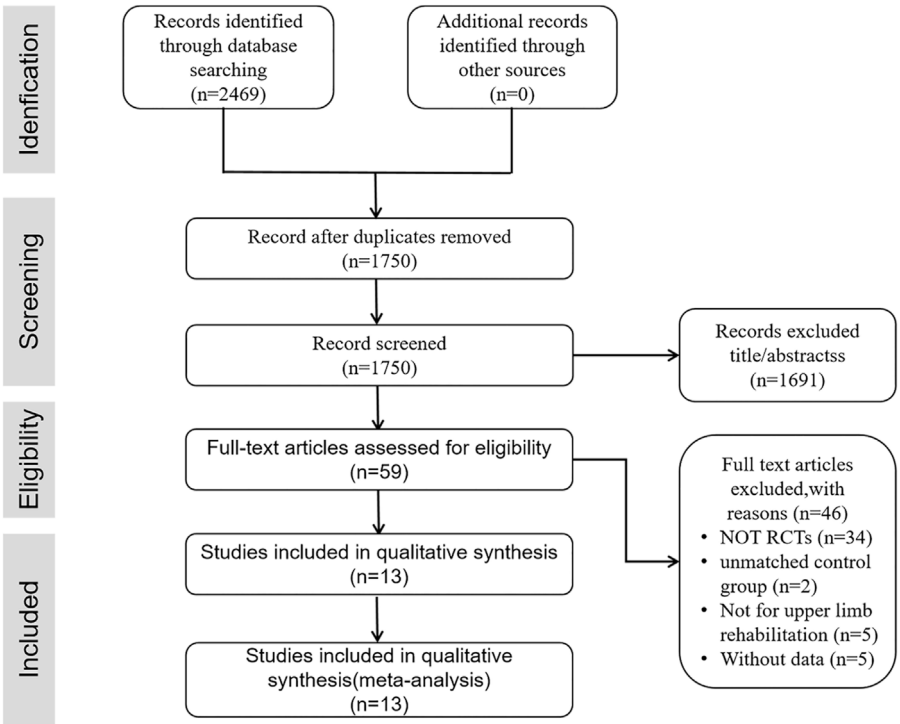

FIGURE 1
The PRISMA flow diagram.

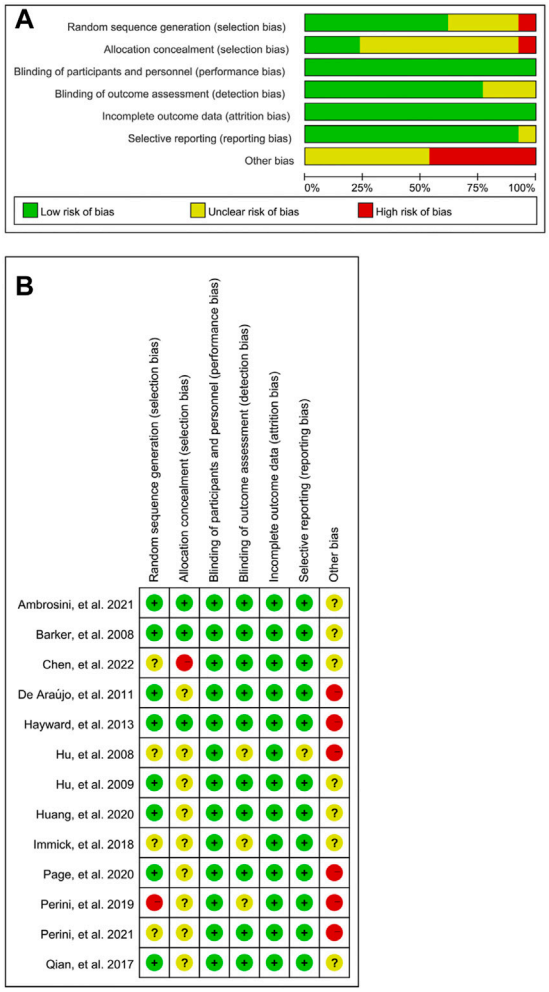

FIGURE 2
Risk of bias in the systematic review. Bias of the included articles is relatively low and stable. **(A)** Risk of bias for all included studies. **(B)** Risk of bias item for each included study.

2.1 Search strategy

We searched studies published before 20 November2022 in these electronic databases: Embase, Scopus, PubMed (MEDLINE), Cochrane library and Web of science. The PICOS (participant, intervention, comparison, outcome and study design) framework was used to research. When determined the systematic keywords to retrieve, we only used P and I to avoid missing crucial articles. The mesh terms used in Embase were "Cerebrovascular Accident" (Participants), "Electromyography" AND "Robotics" (Intervention), and in other databases were related to "Stroke" (Participants), "Electromyography" AND "Robot" (Intervention). Detailed search strategy for those databases could be found in Supplementary Material.

2.2 Inclusion and exclusion criteria

The following was the inclusion criteria: 1) Participants: patients who have been suffering stroke; 2) Intervention: EMG-based robot therapies, including EMG-driven robot, smart rehabilitation systems with EMG, and electromechanical orthosis; 3) Outcomes: measures of upper extremity motor function, spasticity and activity limitation; 4) Study design: randomized controlled, cross-over clinical trials; 5) articles published in English.

The following was the exclusion criteria: 1) studies not focusing on post-stroke patients; 2) no EMG-based robot system described; 3) studies targeting the low limb functions of stroke; 4) case reports; 5) review articles; 6) studies without the full text. Two reviewers assessed each article for inclusion independently. When an identity of views was not reached between the two reviewers, the ratings were discussed among all co-authors until a consensus was reached.

2.3 Outcome measures

Based on the International Classification of Functioning Disability and Health model (ICF) (Reis et al., 2021), we classified outcomes into a) motor control of upper limb, b) muscle tension, and c) activity capacity. The above three aspects were respectively evaluated by Fugl-Meyer Assessment Scale (FMA), modified Ashworth Scale (MAS) and activity limitation (e.g., Box and Blocks Test [BBT], Nine Hole Peg Test, Jebsen-Taylor Hand Function Test, Action Research Arm Test [ARAT]) (Veerbeek et al., 2017; Reis et al., 2021).

2.4 Data extraction

Two authors analyzed the abstracts and contents of each article carefully and extracted the data systematically. If there was a disagreement, it would be resolved by consultation with a third author. We extracted data on the following two aspects: 1) the basic information of the study, including the type of study, demographic characteristics of the subjects, outcomes measures, and 2) information on the EMG-based robot systems, including robot types, parameters of intervention. We contacted authors for original data when partial data were not available.

2.5 Risk of bias assessments

We used the Cochrane risk of bias assessment tool (Higgins and Altman, 2008) to assess each study by two independent authors. This tool contains six items, selection bias, performance bias, detection bias, attrition bias, reporting bias, and other biases (Higgins et al., 2011). Disagreements were resolved by consultation with a third reviewer when necessary.

2.6 Statistical analysis

Review Manager 5.4 (The Nordic Cochrane Centre, The Cochrane Collaboration, Copenhagen, Denmark) was used to analysis. We compared variation in effect sizes on the outcome measures of upper extremity function between the experimental group (EG, using EMG-based robot) and control group (CG, using conventional therapies) before and after intervention.

Then, we used mean difference (MD) or standardized mean difference (SMD) and 95% confidence intervals (CIs) to calculate the pooled effects of outcome measures. In addition, we evaluated heterogeneity by examining forest plots, chi^2 test and I^2 statistic were used to assess the heterogeneity between RCTs. I^2 values range from 0% to 100%, and are considered low at <25%, modest at 25%–50%, and high at >50%, and the t-statistic is being used for the degrees of freedom in the random effects analysis, when the number of studies is small (e.g., <10). Subgroup analyses, meta-regression and sensitivity analysis were computed to evaluate heterogeneity.

3 Results

3.1 Study screening

Figure 1 shows the details of the whole selection process. A total of 2,469 articles were retrieved after searching the databases. And 1750 articles were screened after removing 719 duplicates, of which 1,691 articles were excluded according to the PICOS principle in titles and abstracts. The rest of 59 studies were submit to full-text checking, of which 13 studies were included for meta-analysis.

3.2 Quality of the included studies

Risk of bias for included 13 studies was assessed by two reviewers independently. The results were shown in Figure 2, and sensitive analysis indicated that the results appeared to be stable.

3.3 Study characteristics

Table 1 shows the details of basic information of the included trials. In total, this meta-analysis included 330 subjects (EG, n = 175; CG, n = 155). One study was extracted two experimental groups (Page et al., 2020). For randomized cross-over trial (Chen et al., 2022), data before crossover were extracted. The study with fewest number of participants was carried out by Hayward (n = 8, EG:4, CG:4) (Hayward et al., 2013). The mean duration of post-stroke ranged from 0.87 months to 8.3 years. Most studies concentrated on stroke rehabilitation in chronic stage.

In terms of setting EMG-based robot intervention parameters, the session duration of EMG-based robot ranged from 30 min to 90 min, and the average was 47 min. Eight studies set a session less than 60 min. Three studies (Barker et al., 2008; Hayward et al., 2013; Chen et al., 2022) set a session duration of 60 min, and others were more than 60 min. The average frequency of intervention was 3.79 times/week, and most studies focused on a frequency of 3 times/week. The average intervention period was 6.04 weeks, and most studies continued 4 weeks.

According to the statistics, the effect of the intervention on participants was mainly confirmed by measuring subjects' upper limb motor function, spasticity and activity limitation. The FMA, MAS and assessment for activity level, such as ARAT, BBT were the most commonly used scales.

3.4 Synthesis of results

3.4.1 Effect of EMG-based robot therapy compared with control group

In the subgroup analyses, the EMG-based robot, non EMG-based robot (i.e., conventional rehabilitation robot such as InMotion), and conventional (non-robotic) rehabilitation were compared for the improvement of motor control, spasticity and activity limitation post intervention. Nine studies measured the effect of EMG-based robot on FMA. The results showed that the score of FMA was significantly increased (SMD:0.62, 95% CI:0.29 to 0.95) (Figure 3A). Seven studies focused on MAS, and the results showed MAS changed significantly (MD: −0.42, 95% CI: −0.82 to −0.03) (Figure 3B). Eight studies aimed at activity limitation, the results showed the activity level also improved notably (SMD:0.43, 95% CI:0.05 to 0.82) (Figure 3C). The results of subgroup analyses showed that the effect of EMG-based robot in motor control was superior to both conventional therapies (SMD:0.46, 95% CI:0.03 to 0.89) and robotic therapies (SMD:0.94, 95% CI:0.43 to 1.45) (Figure 4A). In terms of spasticity, no obvious advantage was found for the EMG-based robot over conventional treatment (Figure 4B). The EMG-based robot was more effective than the robotic therapies in the activity limitation (SMD:0.90, 95% CI:0.02 to 1.79) (Figure 4C). However, there is high heterogeneity in the outcome of MAS and activity limitation. The results of meta-regression (Supplementary Figure A1) and sensitivity analysis (Supplementary Figure A2) showed that the sample size contributed to the outcome of MAS and the duration of disease contributed to activity limitation.

3.4.2 The total time of training

As described in previous studies (Wang et al., 2021; Zhang et al., 2022), the amount of intervention was estimated by total time. We discovered that there was a significant difference in the upper limb motor function in both subgroups at the end of treatment between EMG-based robot therapy and conventional therapies, and the effect size was lager in subgroup with total training time ≤1000 min (SMD: 0.67, 95% CI:0.25–1.09) than subset with total time >1000 min (SMD:0.58, 95% CI:0.14–1.03) (Figure 5A). In terms of spasticity, subgroup with total time ≤1000 min had a significant difference between EG and CG (MD: −0.77, 95% CI: −1.06 to −0.48), but no significant difference in the subset with total time >1000 min (MD: −0.02, 95% CI: −0.58 to 0.53) (Figure 5B). Activity limitation also changed significantly in subgroup with total time ≤1000 min (SMD: 0.45, 95% CI:0.06–0.83), rather than subgroup with total time >1000 min (SMD:0.44, 95% CI: −0.58–1.45) (Figure 5C).

3.4.3 The intervention mode

The intervention modes provided by EMG-based robot included EMG-driven robot with electrical stimulation (ES) and the type without ES. Subgroup analysis showed that the robot type with ES (SMD: 0.91, 95% CI:0.44–1.37) had larger effect size on the FMA than the type without ES (SMD: 0.42, 95% CI: 0.01–0.82) at the end of treatment (Figure 6A). The outcome of MAS at the end of treatment showed no significant difference between EG and CG in both subgroup (Figure 6B). In terms of the activity limitation, we found that the subgroup with ES (SMD:0.60, 95% CI:0.17–1.04) was better than the subgroup without ES (SMD: −0.06, 95% CI: −0.62–0.49) (Figure 6C).

TABLE 1 Basic information about included studies.

Study	Intervention EG vs. CG	Participants	Age(years)	Post-stroke duration(months)	Intervention time setting			Outcome measures
			Mean ± SD	Mean ± SD	Session duration (minutes)	Frequency (weeks)	Weeks	
Ambrosini et al., 2021	EMG-driven FES robot vs. Conventional Therapies	EG:36	EG:60.9 (13.7)	EG:2.11 (1.96)	30	3	9	ARAT, BBT
		CG:36	CG:67.8 (12.2)	CG:2.07 (2.56)				
Barker et al. (2008)	Smart rehabilitation system and stim vs. smart rehabilitation system alone	EG:10	EG:61 (16)	EG:60 (58.8)	60	3	4	MAS
		CG;13	CG:67 (8)	CG:40.8 (31.2)				
Chen et al. (2022)	EMG-driven robot vs. Task Oriented	EG:14	EG:54.58 (10.98)	EG:37.07 (34.39)	60	3	8	FMA, ARAR
		CG:10	CG: 64.98 (8.22)	CG:59.8 (43.34)				
De Araújo et al., 2011	Electromechanical orthosis vs. Conventional Therapies	EG:6	EG:42.83 (14.04)	EG:21.67 (11.83)	50	3	8	FMA, MAS
		CG:6	CG:52.67 (17.84)	CG:19.00 (11.01)				
Hayward et al. (2013)	Smart rehabilitation system and stim vs. smart rehabilitation system alone	EG:4	EG:69 (10)	EG:1.53 (0.4)	60	5	4	MAS
		CG:4	CG:56 (24)	CG:0.87 (0.2)				
Hu et al. (2009)	EMG-Driven robot vs. passive device	EG:15	EG:49.2 (14.7)	EG:56.4 (50.4)	36.5	3–5	4–7	FMA, MAS, ARAT
		CG:12	CG:53.3 (10.4)	CG:61.2 (49.2)				
Hu et al. (2009)	EMG-Driven robot vs. passive device	EG:5	EG:50.2 (10.2)	EG:12	36.5	3–5	4–6	FMA, MAS
		CG:5	CG:50.2 (10.2)	CG:12				
Huang et al., 2020	EMG-Driven NMES-robot vs. robot	EG:15	EG: 57.33 (9.19)	EG:99.24 (51.84)	60	3–5	4–7	FMA, MAS, ARAT
		CG:15	CG: 60.07 (6.88)	CG:74.4 (40.92)				
Immick et al., 2018	EMG-driven robot vs.Conventional Therapies	EG:19	EG:59.0 (15.9)	EG:2.14 (2.2)	30	3	9	ARAT, BBT
		CG:20	CG:67.7 (12.1)	CG:2.65 (3.13)				
Page et al. (2020)	Myomo vs. RTP	EG:14	EG: 55.79 (9.25)	EG: N	30	3	8	FMA
		CG: 5	CG: 57.22 (7.68)	CG: N				
Page et al. (2020)	Myomo + RTP vs. RTP	EG:8	EG:52.89 (11.38)	EG: N	30	3	8	FMA
		CG:5	CG: 57.22 (7.68)	CG: N				
Perini et al., 2019	MeCFES + robot vs. Conventional Therapies	EG:6	EG:65.5 (23.1)	EG:20 (10.3)	45	5	4	FMA, ARAT
		CG:5	CG:65.5 (23.1)	CG:20 (10.3)				
Perini et al., 2021	MeCFES + robot vs. Task Oriented	EG:9	EG:58.7 (20.6)	EG:35.3 (44.5)	90	5	4	FMA, BBT
		CG:9	CG:61.4(9)	CG: 42 (44.7)				
Qian et al. (2017)	EMG-Driven NMES-robot vs. Conventional Therapies	EG:14	EG:54.6 (11.3)	EG:0.83	40	5	4	FMA, MAS, ARAT
		CG:10	CG:64.6 (3.43)	CG: 0.46				

EG, experimental group; CG, control group; Myomo, myoelectric device; RTP, repetitive, task-specific practice; NMES, neuromuscular electrical Stimulation; MeCFES, myoelectric control functional electrical stimulation; FMA, fugl-mayer assessment; MAS, modified ashworth scale; ARAT, action research arm test; BBT, box and block test.

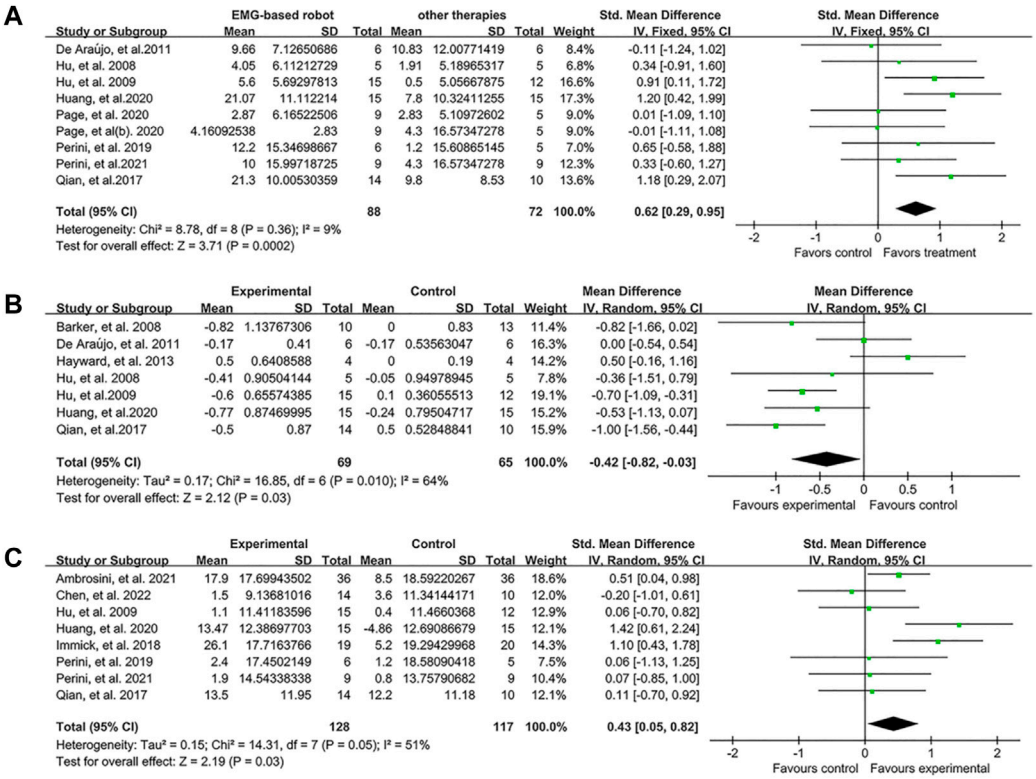

FIGURE 3
Forest plot analysis of the effect of EMG-based robot therapy vs. conventional therapies. **(A)** Forest plot analysis of the effect of EMG-based robot therapy vs. conventional therapies on FMA. **(B)** Forest plot analysis of the effect of EMG-based robot therapy vs. conventional therapies on MAS. **(C)** Forest plot analysis of the effect of EMG-based robot therapy vs. conventional therapies on activity limitation.

3.4.4 Stage of stroke

The stage of stroke was evaluated according to the duration after the onset time, and the participants were classified into subacute (≤6m) and chronic (>6m) group (Wang et al., 2021). For upper limb motor function, most studies focused on chronic stage and just one study designed by Qian, et al. (Qian et al., 2017) was in subacute stage. Analysis showed that FMA score was significantly changed in chronic group (SMD:0.55, 95% CI:0.23–0.88) and subacute group (SMD:1.18, 95% CI:0.29–2.07) (Figure 7A). For spasticity, there were no significant difference between two groups (Figure 7B). When comes to the activity limitation, the result showed the effect was better in subacute group (SMD:0.60, 95% CI:0.10–1.09) than chronic group (SMD:0.30, 95% CI: −0.31–0.91) (Figure 7C).

4 Discussion

In the present meta-analysis, a total of 12 RCTs and one cross-over clinical trial were analyzed, which included a total of 330 subjects (EG, $n = 175$; CG, $n = 155$). The results indicated that EMG-based robot was more effective than conventional therapies, including physical therapy, occupational therapy, passive training, and conventional robotic therapy. Results from subgroup analyses further revealed that the efficacy of the treatment was better in patients in the subacute stage, those who received a total treatment time of less than 1000 min, and those who received electromyography (EMG)-based robotic therapy combined with electrical stimulation (ES). These findings suggest that EMG-based robot therapy may be considered a promising rehabilitation method to improve upper limb dysfunction in post-stroke patients.

4.1 Effect of EMG-based robot

The findings presented in Figures 3, 4 demonstrate that EMG-based robot therapy can significantly enhance upper limb motor function, reduce spasticity, and improve activity level in post-stroke patients, as compared to the robotic therapies and conventional therapies. Following a stroke occurrence, the injured brain undergoes a reorganization process that involves recruiting pathways homologous to the damaged regions in function but distinct in anatomy, synaptogenesis, dendritic arborization, and reinforcing synaptic connections (Rossini et al., 2003). Such adaptive brain changes seem to be closely related to rehabilitation and motor training, contributing to the improvement of functional outcomes (Chen et al., 2018; Sampaio-Baptista et al., 2018; Sheng et al., 2022). High-intensity repetitive training is one of the principles of post-stroke rehabilitation (Langhorne et al., 2011). Although there are no clear guidelines for the optimal level of practice, it is widely accepted that more intensive training is beneficial. Robot-assisted therapy can provide patients with repetitive, high-intensity training and improve their motor function (Zhang et al., 2017; Kim et al., 2020; Wang et al., 2021; Yeung et al.,

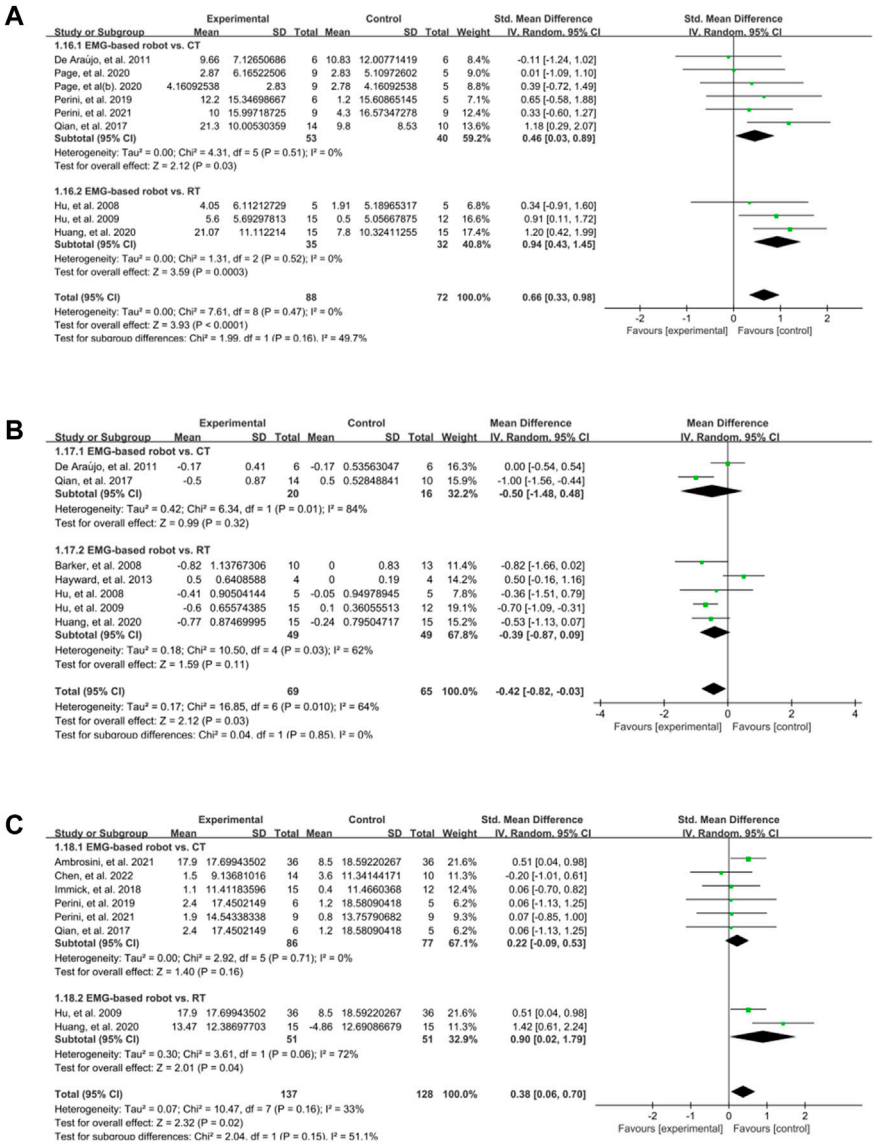

FIGURE 4
A subgroup analysis of the effect of EMG-based robot vs. different control groups on outcomes. **(A)** Forest plot analysis of the effects of different control groups on FMA. **(B)** Forest plot analysis of the effects of different control groups on MAS. **(C)** Forest plot analysis of the effects of different control groups on activity limitation.

2021). The possible mechanism involves enhancing neural plasticity and neuronal activity to facilitate neuroplasticity change (Chen et al., 2019; Xing and Bai, 2020). According to the Hebbian learning rule, the connections between neurons are strengthened when neurons are simultaneously active (that is, long-term potentiation) (Orbach, 1998). In addition, signals from EMG-based robot can reflect the voluntary movement intention of patients in real time (Rong et al., 2015; Zhuang et al., 2021), which may increase active patient participation and promote interaction between humans and machines. Recording of muscle activity by EMG has proved to be helpful to explore the activity state of muscle tissue and the control mechanism of the nervous system under different task states after stroke (Ma et al., 2017; Chen et al., 2018), and then be useful for evaluating central and peripheral determinants of motor dysfunction which facilitates the understanding of mechanism behind rehabilitation intervention (Sheng et al., 2022; Li et al., 2023). Voluntary motor intention is crucial in the rehabilitation of motor function after stroke (Yang et al., 2022). Rehabilitation training that incorporates participants' intention input is more effective which further facilitate active participation (Hu et al., 2009; Hu et al., 2021; Zhang et al., 2022) and interactive control (Cozens, 1999). The formation of a sensorimotor cycle by the voluntary intention output and the afferent sensor might facilitate motor relearning in post-stroke (Cauraugh et al., 2000). Furthermore, the effectiveness of repeated training increased substantially with the increase interaction between patients and machine (Hu et al., 2009; Hu et al., 2021). In general, the results of current study support the clinical application of EMG-based robot therapy in improving upper limb dysfunction in patients after

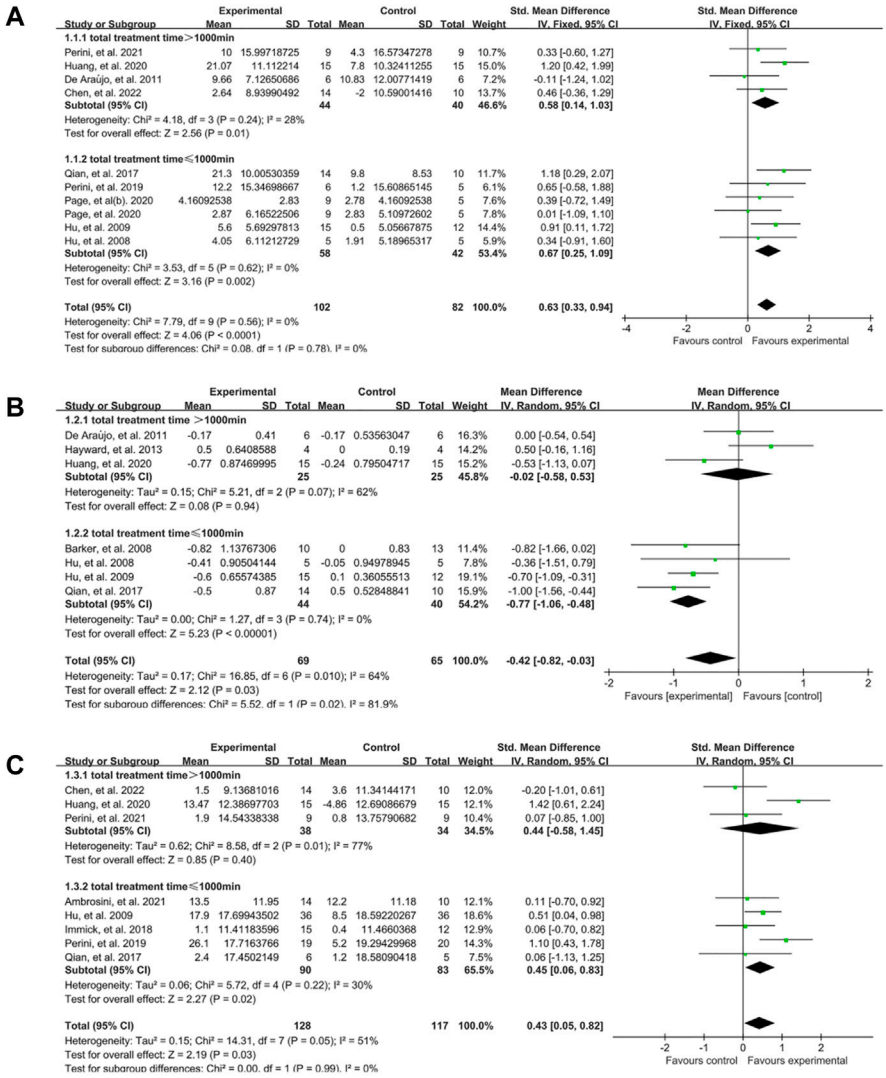

FIGURE 5
A subgroup analysis of the effect of EMG-based robot with different total training time vs. conventional therapies on outcomes. **(A)** Forest plot analysis of the effects of different total time of training on FMA. **(B)** Forest plot analysis of the effects of different total time of training on MAS. **(C)** Forest plot analysis of the effects of different total time of training on activity limitation.

stroke. However, more investigation is needed to reveal the relation between the changes of brain function and improvements of neuromuscular systems (i.e., by using cortical-muscular coherence technique) during the EMG-based robot training. This might be helpful for the understanding of the potential mechanism related to neuroplasticity in post-stroke patients.

4.2 The training intensity and mode of the EMG-based robotic intervention

The subgroup analysis suggested that the effect of the subgroup with total treatment time ≤ 1000 min was better than the subgroup with total time >1000 min (Figure 5), indicating the optimal treatment parameters to achieve the best effect of EMG-based robot therapy remains to be determined. Post-stroke fatigue is a common complication that negatively impacts patient's rehabilitation outcomes (Duncan et al., 2012; Finsterer and Mahjoub, 2014; Maaijwee et al., 2015). One prevalent hypothesis is that physical deconditioning may contribute to fatigue following stroke (Duncan et al., 2012). One study focused on brain-computer interface (BCI) indicated that mental fatigue may also play a role in poorer BCI performance (Foong et al., 2020). Fatigue may affect patient motivation during training and interaction between human and machine, thereby hampering the efficiency of EMG-based robot. However, the relationship between post-stroke fatigue and motor training is rarely studied, and further research is needed to explore this aspect.

In the context of training mode, our results showed that the efficiency of the EMG-based robot with electrical stimulation (ES) was superior to the type without ES (Figure 6). Clinically, ES can be used to activate muscles, prevent muscle atrophy, and increase muscle

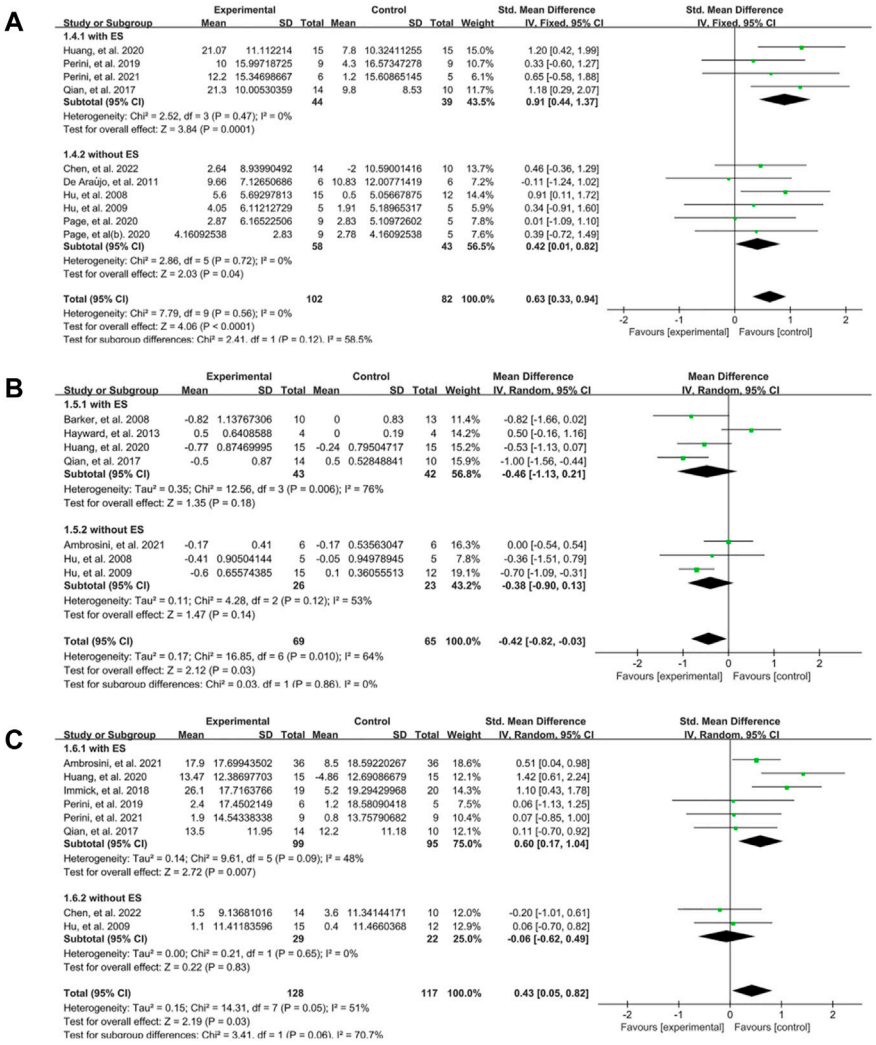

FIGURE 6
A subgroup analysis of the effect of different EMG-based robots vs. conventional therapies on outcomes. **(A)** Forest plot analysis of the effects of different intervention mode on FMA. **(B)** Forest plot analysis of the effects of different intervention mode on MAS. **(C)** Forest plot analysis of the effects of different intervention mode on activity limitation.

strength (Hu et al., 2021; Li et al., 2022). In addition, sensory dysfunction is a common complication after stroke (Sullivan and Hedman, 2008; Tyson et al., 2008), which is related to the reduction of motor function recovery (Kusoffsky et al., 1982). ES may effectively improve sensory awareness of paralyzed muscles after stroke (Mäenpää et al., 2004), and elicit sensory feedback to the cortex during muscle contraction to facilitate motor relearning (Sujith, 2008). The sensory feedback is beneficial for motor function recovery post stroke (Sharififar et al., 2018). In the subgroup that received ES, the sensory feedback from the ES to the affected limb may be beneficial in motor function improvement. But excessive ES might impede the effect of motor training (Chae et al., 2002) and it is crucial to explore the optimal proportion of assistance from both ES and robot (Li et al., 2022). Hu et al., (Hu et al., 2011), reported that the performance of wrist tracking could be better with the 1:1 assistance from both ES and robot. But the small sample size and lack of long-term testing limited the findings of this study.

The subgroup analysis also showed that the EMG-based robot was more effective in subacute stage than chronic stage (Figure 7), which is consistent with previous studies (Mehrholz et al., 2018; Dehem et al., 2019; Dromerick et al., 2021). Early participation in rehabilitation is crucial for motor recovery as it can facilitate brain reorganization, optimize motor responsiveness and spontaneous neural plasticity, which may contribute to better rehabilitation outcomes in post-stroke patients, (Zeiler and Krakauer, 2013; Ng et al., 2015). Improved motor function in the subacute phase is more likely to generalize into activities of daily living (Flöel et al., 2014; Tomori et al., 2015).

4.3 Study limitations

While our subgroup analysis indicated that the effect of EMG-based robot was better in patients in the subacute stage, those who received a total treatment time of less than 1000 min, and those who received

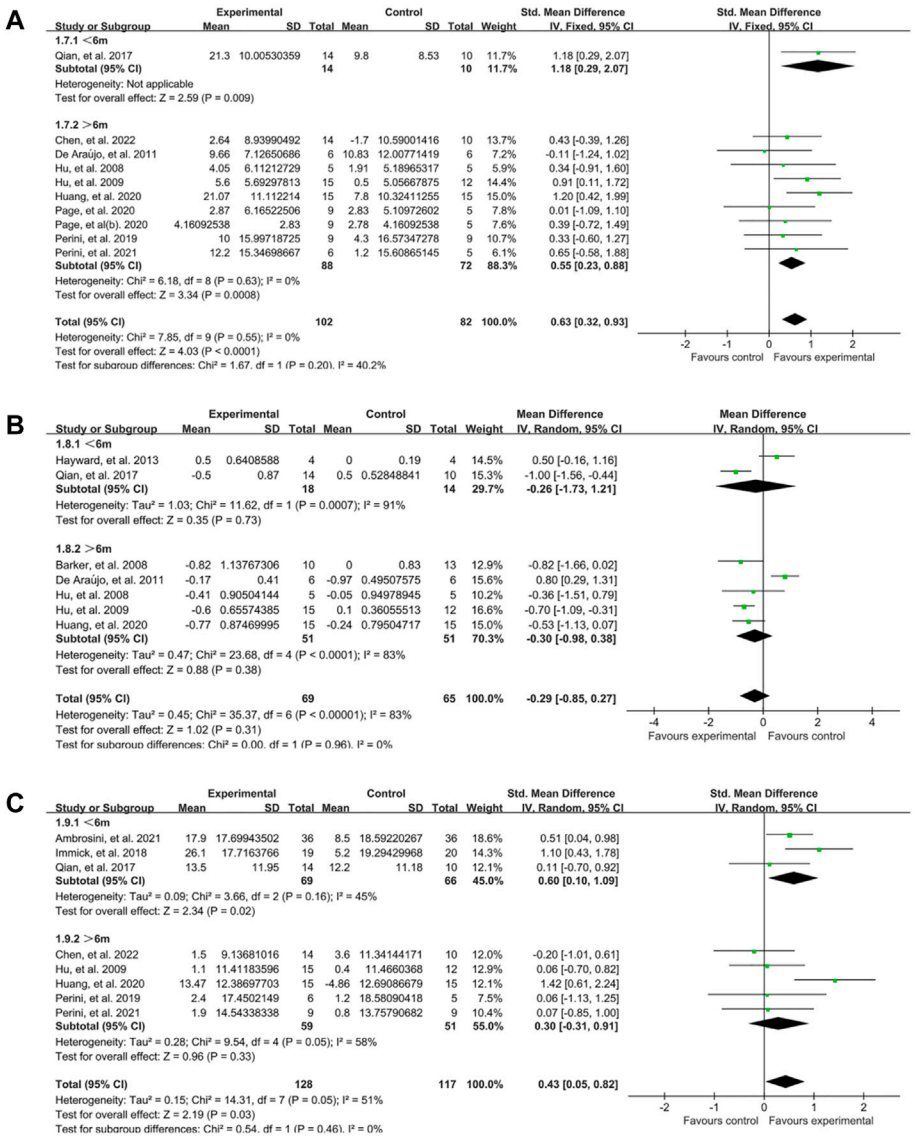

FIGURE 7
A subgroup analysis of the effect of EMG-based robot for different stage of stroke. **(A)** Forest plot analysis of the effects of different stage of stroke on FMA. **(B)** Forest plot analysis of the effects of stage of stroke on MAS. **(C)** Forest plot analysis of the effects of stage of stroke on activity limitation.

EMG-based robot combined with ES, it is important to acknowledge the limited sample size of the included studies and the potential impact on the validity of the results. Furthermore, the language restriction of our study to English-language articles may have introduced a selection bias which may also limit the generalizability of the results. The EMG-based robot requires active training and may not be suitable for patients with cognitive impairment. Future studies may explore the relationship between post-stroke fatigue and functional training to determine the best parameters of the EMG-based robot for the upper limb function, which could ultimately results in improved clinical outcomes for stroke survivors.

5 Conclusion

The present study provides evidence that EMG-based robot therapy is superior to conventional therapies in improving upper limb motor control, spasticity, and activity limitation in post-stroke patients. These findings suggest that EMG-based robot therapy could be a promising rehabilitation method for promoting the recovery of upper extremity function in this patient population. Further research should explore optimal parameters of EMG-based robot therapy and its long-term effects on upper limb function in post-stroke patients.

Data availability statement

The original contributions presented in the study are included in the article/supplementary materials, further inquiries can be directed to the corresponding authors.

Author contributions

LL, HH, and YH: conceptualization. YH, XW, and LL: methodology. YH and LL: writing–original draft preparation. YH, XW, WZ, HH, and LL: writing–review and editing. All authors contributed to the article and approved the submitted version.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Supplementary material

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fphys.2023.1172958/full#supplementary-material

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Levels of ACL-straining activities increased in the six months prior to non-contact ACL injury in a retrospective survey: evidence consistent with ACL fatigue failure

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Introduction: Recent evidence has emerged suggesting that a non-contact anterior cruciate ligament (ACL) tear can result from repetitive submaximal loading of the ligament. In other words, when the intensity of ACL-straining athletic activities is increased too rapidly, microdamage can accumulate in the ligament beyond the rate at which it can be repaired, thereby leading to material fatigue in the ligament and its eventual failure. The objective of this survey-based exploratory study was to retrospectively determine whether the levels of various athletic activities performed by ACL-injured patients significantly changed during the 6 months before injury.

Methods: Forty-eight ACL-injured patients completed a survey to characterize their participation in various activities (weightlifting, sport-specific drills, running, jumping, cutting, pivoting/twisting, and decelerating) at three timepoints (1 week, 3 months, 6 months) prior to ACL injury. Activity scores, which summarized the frequency and intensity of each activity, were calculated for each patient at each time interval. A series of linear mixed-effects regression models was used to test whether there was a significant change in levels of the various activities in the 6-month period leading up to ACL injury.

Results: Patients who sustained a non-contact ACL injury markedly increased their sport-specific drills activity levels in the time leading up to injury ($p = 0.098$), while those patients who sustained a contact ACL injury exhibited no change in this activity during the same time period ($p = 0.829$). Levels of running, jumping, cutting, pivoting/twisting, and decelerating increased for non-contact ACL-injured patients but decreased for contact ACL-injured patients, though not significantly (p values > 0.10). Weightlifting activity significantly decreased leading up to injury among contact ACL-injured patients ($p = 0.002$).

Discussion: We conclude that levels of ACL-straining athletic activities or maneuvers in non-contact ACL-injured patients markedly increased in the 6 months leading up to their injury, providing evidence that changing levels of certain activities or maneuvers may play a role in ACL injury risk. This warrants further investigation of the hypothesis that too rapid an increase in activities or

maneuvers known to place large loads on the ACL can cause microdamage to accumulate in the ligament, thereby leading to failure.

KEYWORDS

anterior cruciate ligament, injury, physical activity, fatigue, microdamage, sports

1 Introduction

Recently, the prevailing dogma that non-contact anterior cruciate ligament (ACL) injuries are the result of a single catastrophic maneuver that overloads the healthy ACL has been challenged. While this can happen on occasion, evidence is mounting for a second injury mechanism: namely, that of "tissue fatigue", by which repetitive submaximal loading of the ACL can cause microdamage to accumulate in the ligament, thereby weakening it until it causes the ACL to fail (Lipps et al., 2013; Wojtys et al., 2016; Chen et al., 2019; Kim et al., 2022). For instance, in an *in vitro* simulation of repetitive pivot-landings known to place the ACL under marked strain, the ACLs of cadaveric knees failed after fewer than 100 of those submaximal loading cycles (Lipps et al., 2013). Additionally, ACL tissue microdamage found in another group of cadaveric knees subjected to these repetitive pivot-landings (Kim et al., 2022) proved to be similar to that found to have accumulated in *ex vivo* ACL collagen fibrils and fibers of non-contact ACL injury patients at the time of surgery (Chen et al., 2019). Therefore, the ACL may fail when the accumulation of microdamage outpaces repair (i.e., catabolic state), similar to ulnar collateral ligament (UCL) injuries in baseball pitchers. In these athletes, it has long been established that repetitive throwing at high velocities without proper recovery can lead to an UCL tear (Mirowitz and London, 1992). For this reason, the total volume and frequency of pitches are monitored and limited in baseball. With regard to the ACL, however, there is as yet no such *in vivo* evidence that it can fail via repetitive sub-maximal loading. The focus of this study, therefore, is to address this knowledge gap by testing our overall hypothesis that if the ACL can fail via a 'tissue fatigue' injury mechanism, we would expect the frequency and/or intensity of activities/maneuvers that significantly strain the ACL (hereafter referred to as "ACL-straining athletic activities or maneuvers") to have increased substantially prior to a patient's ACL non-contact injury, before the ACL has had adequate time to adapt. In such cases, natural repair homeostasis would be replaced by a catabolic state where the rate of ACL tissue degradation exceeds that of synthesis, leading to an eventual ligament failure.

The purpose of this exploratory study was to determine whether the frequency and/or intensity at which ACL-injured patients performed various athletic activities or maneuvers known to markedly strain the ACL changed in the 6-month period leading up to their ACL injury. We selected a 6-month period because we estimated that given the ACL's relatively slow repair rate (Rucklidge et al., 1992), it would not be able to adapt to a substantial increase in ACL-straining athletic activities or maneuvers within this timeframe. Although the exact repair rate of the ACL is unknown, we do know that ligaments heal more slowly than other tissues (Panjabi, 2006; Jung et al., 2009) given that ligaments have lower metabolic activity rates than muscle, bone, or cartilage (Robi et al., 2013; Nyland et al., 2022). In the present study, ACL reconstruction patients completed a questionnaire to retrospectively quantify the frequency and intensity of their athletic activity/maneuver leading up to injury. The questionnaire was designed to collect data on various athletic activities and maneuvers, in particular those that are known to significantly strain the ACL and have been associated with ACL injuries. Through video analysis for instance, non-contact ACL injuries have been found to occur while cutting/changing direction, pivoting/twisting, decelerating, and landing from a jump (Olsen et al., 2004; Krosshaug et al., 2007; Boden et al., 2009; Walden et al., 2015; Della Villa et al., 2020). We hypothesized that the frequency/intensity of ACL-straining athletic activities or maneuvers (i.e., decelerating, jumping, cutting, pivoting/twisting, sport-specific drills) would markedly increase in the 6-month period leading up to ACL injury, especially among those patients that sustained a non-contact injury.

2 Materials and methods

This study was a retrospective case series survey study. Patients having suffered an ACL injury were asked to complete one questionnaire post-injury to quantify the frequency and intensity at which they performed various athletic activities/maneuvers in the 6-month period leading up to their injury. In particular, it retrospectively quantified these data for the 6-month, 3-month, and 1-week pre-injury timepoints.

2.1 Participants

We recruited a convenience sample of 48 ACL-injured patients (12 males/36 females; age: 19.1 ± 6.5 years; height†: 1.7 ± 0.1 m; body mass†: 67.3 ± 12.7 kg; BMI†: 23.6 ± 4.0 kg/m^2; †data based on 47 patients due to missing data) who voluntarily completed a questionnaire at their initial post-injury visit (on average†, 56.8 ± 118.7 days post-injury) to determine their activity patterns leading up to their ACL injury. Most of the patients were athletes (46/48, 96%; playing experience: 11.4 ± 7.7 years), with most athletes participating in sports deemed risky for ACL injury, such as soccer and basketball. Many participants were also multi-sport athletes (Table 1). All ACL injuries were confirmed by physician examination and magnetic resonance imaging (MRI) evaluation. Fifteen (6 male/9 female) patients suffered a contact ACL injury, meanwhile 33 (6 male/27 female) patients suffered a non-contact injury. We excluded patients with a previous injury to the ACL and/or meniscus of their ipsilateral knee. All qualifying patients of the senior author at MedSport, the sport medicine clinic of the University of Michigan who sustained a primary ACL injury from 2016 to 2019 were invited to participate in this study. This study

TABLE 1 Frequency of Sports in which Patients Participated Prior to ACL Injury.

Sport	Frequency total # (M/F) % of total patients (% of M/F)
Multi-sport	18 (6/12) 37.5% (50.0/33.3)
Ball sports	
Soccer	17 (3/14) 35.4% (25.0/38.9)
Basketball	12 (5/7) 25.0% (41.7/19.4)
Volleyball	6 (0/6) 12.5% (0/16.7)
Lacrosse	7 (0/7) 14.6% (0/19.4)
Football	5 (5/0) 10.4% (41.7/0)
Softball	4 (0/4) 8.3% (0/11.1)
Baseball	1 (1/0) 2.1% (8.3/0)
Rugby	1 (0/1) 2.1% (0/2.8)
Golf	1 (0/1) 2.1% (0/2.8)
Combined discipline/running/cycling	
Track & field	4 (2/2) 8.3% (16.7/5.6)
Dance	3 (0/3) 6.3% (0/8.3)
Cross country/running	2 (0/2) 4.2% (0/5.6)
Biking	1 (1/0) 2.1% (8.3/0)
Snow Sports	
Skiing	3 (0/3) 6.3% (0/8.3)
Snowboarding	2 (1/1) 4.2% (8.3/2.8)
Combat sports	
Wrestling	2 (2/0) 4.2% (16.7/0)
Mixed martial arts/Brazilian jiu-jitsu	1 (0/1) 2.1% (0/2.8)
Water Sports	
Swimming	1 (0/1) 2.1% (0/2.8)
Water polo	1 (1/0) 2.1% (8.3/0)
Rowing	1 (0/1) 2.1% (0/2.8)
No Sport	2 (0/2) 4.2% (0/5.6)

M: male patients; F: female patients.

(HUM00109196) was approved by the Institutional Review Board of the University of Michigan Medical School (IRB00001996).

2.2 Sport and Physical Activity Questionnaire

Each participant completed a "Sport & Physical Activity Questionnaire" we developed, which retrospectively assessed the level of various athletic activities and maneuvers of the participant in the 6-month period prior to their ACL injury (see Supplemental Material file). We divided the questionnaire into three identical sections, each corresponding to a different time point prior to injury: 6 months, 3 months, and 1 week. Within each section, we asked patients questions about their activity patterns for seven different activities: weightlifting, sport-specific drills, running, jumping, cutting, pivoting/twisting, and decelerating. They described the nature of their participation relating to each activity by three measures: 1) indicating (Yes or No) whether they participated in the activity; 2) providing the frequency (minutes per day and days per week) at which they participated in the activity; and 3) rating the intensity at which they participated in the activity on a 0–10 scale (0 = Not intense; 10 = Most intense).

The athletic activities or maneuvers of interest in the questionnaire were not mutually exclusive. In other words, it was possible for a participant to simultaneously engage in several of the activities or maneuvers during a single sporting/exercise session. This is especially true for the many patients that participated in sports that utilize several of the activities of interest. For example, a patient that participated in two 90-min soccer practices and one 30-min running bout per week could have a total frequency of 210 min/week for the "running" activity and could have a total frequency of 180 min/week for the "sport-specific drills", "jumping," "cutting," "pivoting/twisting," and "decelerating" activities (i.e., soccer typically includes the activities of running, sport-specific drills, jumping, cutting, pivoting/twisting, and decelerating).

2.3 Quantitative analysis

For analysis purposes, we categorized ACL-straining athletic activities or maneuvers (cutting, jumping, pivoting/twisting, decelerating, and sport-specific drills) as '*risky*'. Sports-specific

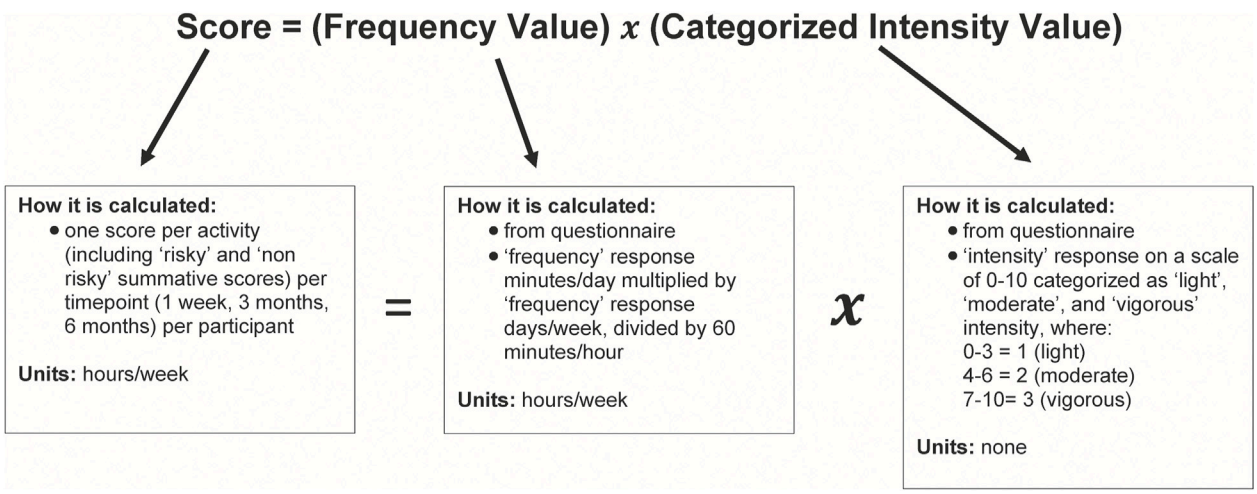

FIGURE 1
Activity score calculation. Diagram illustrating the calculation of activity scores. Diagram describes each component of the score calculation: the score, frequency value, and categorized intensity value. We retrieved the frequency value and categorized intensity value from patient questionnaires.

drills were categorized as "risky" because nearly all the participants practiced a sport that included a considerable amount of cutting, jumping, pivoting/twisting, or decelerating in the sports' basic maneuvers. We categorized activities or maneuvers not deemed risky for ACL injury (weightlifting and running) as "*non-risky*". We summarized the frequency and intensity at which the participants partook in each activity or maneuver by a single score for each timepoint for each participant. We calculated this score using a custom formula that was created in consultation with an experienced statistician with our institution's statistical consulting services. The formula multiplies the participants' responses for frequency (calculated in 'hours per week') and a categorized intensity value based on their intensity response (Figure 1). The intensity values were categorized as 'light', 'moderate', and 'vigorous' intensity. The categorization was as follows: 1 for "light" intensity (patient response values 0–3); 2 for 'moderate' intensity (values 4–6); and 3 for "vigorous" intensity (values 7–10). The grouping of questionnaire intensity values was selected to parallel those of the Physical Activity Guidelines for Americans (Olson et al., 2018), which categorizes intensity values of 5–6 (on a 1–10 scale) as moderate intensity and 7–8 as the minimum for vigorous intensity. Additionally, we created summative "risky" and "non-risky" scores—the sum of the scores of all activities or maneuvers deemed "risky" and "non-risky", respectively, in terms of ACL injury risk as listed above—for each participant at each timepoint.

2.4 Statistical analysis

We used a series of linear mixed-effects regression models to test whether there was a significant change in the levels of the aforementioned activities or maneuvers in the 6-month period leading up to ACL injury. The outcome variable for each model was the activity *score* for each activity or maneuver type (i.e., weightlifting, sport-specific drills, running, jumping, cutting, pivoting/twisting, decelerating, all "risky" activities or maneuvers, and all "non-risky" activities or maneuvers). The predictor variables for all models were *time*, *injury type*, the interaction of *time x injury type* and *patient*. The variables *time* and *injury type* were treated as fixed effects, where *time* was a continuous variable representing the number of days relative to injury (6 months = −182.5 days; 3 months = −91.25 days; 1 week = −7 days; injury = 0 days) and *injury type* was a categorical variable (coded as 0 = non-contact and 1 = contact ACL injury). We treated *patient* as a random effect. In addition, *post hoc* analyses were performed using data obtained from the regression models. First, we statistically compared the "time vs. score" slope of each injury type group to zero to test whether there was a significant change in levels of certain activities or maneuvers in the 6-month period leading up to ACL injury in each group. Second, we calculated the estimated marginal means of the activity scores for the three times points for each activity or maneuver type. These data were used for graphical purposes. Data were analyzed in R version 3.6.3 (R Core Team, 2020). The linear mixed-effects models were created with the lmer() function of the lme4 package (Bates et al., 2015). The statistical significance of the slopes were tested using the emtrends() functions of the emmeans package (Lenth et al., 2020). Lastly, the estimated marginal means were calculated with the emmeans() function also of the emmeans package. We set the alpha level for statistical significance at 0.10.

3 Results

In those patients that identified themselves as athletes, most ACL injuries occurred during their season (33/46; 72%), with fewer injuries occurring in the preseason (7/46;15%) and offseason (5/46;11%); one patient did not provide this information. Of all the patients (female/male patients), 66.7% (F: 61.1%/M: 83.3%), 89.1% (F: 88.2%/M: 91.7%), 91.1% (F:

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TABLE 2 Statistics of the Linear Mixed-Effects Regression Models.

Variable	EMS/β[a]	SE[a]	p value[a]	L_95%CI[a]	U_95%CI[a]
Weightlifting					
Time—all patients	−0.016	0.005	**0.003**	−0.026	−0.005
Time—noncontact injuries	−0.003	0.006	0.542	−0.015	0.008
Time—contact injuries	−0.028	0.009	**0.002**	−0.046	−0.011
Time*injury type	−0.025	0.010	**0.020**	−0.045	−0.004
Sport-specific drills					
Time—all patients	0.016	0.016	0.316	−0.016	0.048
Time—noncontact injuries	0.026	0.016	**0.098**	−0.005	0.057
Time—contact injuries	0.006	0.028	0.829	−0.049	0.061
Time*injury type	−0.020	0.032	0.529	−0.082	0.042
Running					
Time—all patients	−0.007	0.011	0.540	−0.029	0.016
Time—noncontact injuries	0.011	0.012	0.378	−0.013	0.034
Time—contact injuries	−0.024	0.019	0.208	−0.062	0.014
Time*injury type	−0.035	0.023	0.126	−0.079	0.009
Jumping					
Time—all patients	−0.010	0.011	0.375	−0.033	0.013
Time—noncontact injuries	0.011	0.011	0.347	−0.012	0.034
Time—contact injuries	−0.031	0.020	0.119	−0.070	0.008
Time*injury type	−0.042	0.023	**0.070**	−0.087	0.003
Cutting					
Time—all patients	−0.005	0.011	0.652	−0.026	0.017
Time—noncontact injuries	0.001	0.012	0.929	−0.022	0.024
Time—contact injuries	−0.011	0.018	0.552	−0.047	0.025
Time*injury type	−0.012	0.021	0.583	−0.054	0.030
Pivoting/Twisting					
Time—all patients	−0.007	0.010	0.478	−0.026	0.012
Time—noncontact injuries	0.012	0.010	0.237	−0.008	0.033
Time—contact injuries	−0.026	0.016	0.115	−0.059	0.007
Time*injury type	−0.038	0.019	**0.051**	−0.076	−0.000
Decelerating					
Time—all patients	−0.004	0.010	0.693	−0.024	0.016
Time—noncontact injuries	0.008	0.011	0.471	−0.014	0.029
Time—contact injuries	−0.016	0.017	0.349	−0.049	0.018
Time*injury type	−0.024	0.020	0.239	−0.063	0.015
Non-risky activities					
Time—all patients	−0.024	0.014	**0.093**	−0.051	0.004
Time—noncontact injuries	0.011	0.014	0.441	−0.017	0.040
Time—contact injuries	−0.058	0.024	**0.016**	−0.105	−0.011
Time*injury type	−0.069	0.028	**0.014**	−0.123	−0.014
Risky activities					
Time—all patients	0.017	0.063	0.784	−0.108	0.143
Time—noncontact injuries	0.073	0.061	0.236	−0.049	0.194
Time—contact injuries	−0.038	0.110	0.732	−0.258	0.182
Time*injury type	−0.111	0.125	0.381	−0.354	0.138

Abbreviations: EMS, estimated marginal slope; SE, standard error; L_95%CI: lower limit of 95% confidence interval; U_95%CI: upper limit of 95% confidence interval; units of "time" are days; bold values indicate statistical significance ($p < 0.10$).[a] refers to the estimated marginal slope data for the "time" variables and to the regression coefficient (β) data for the "time*injury type" interaction.

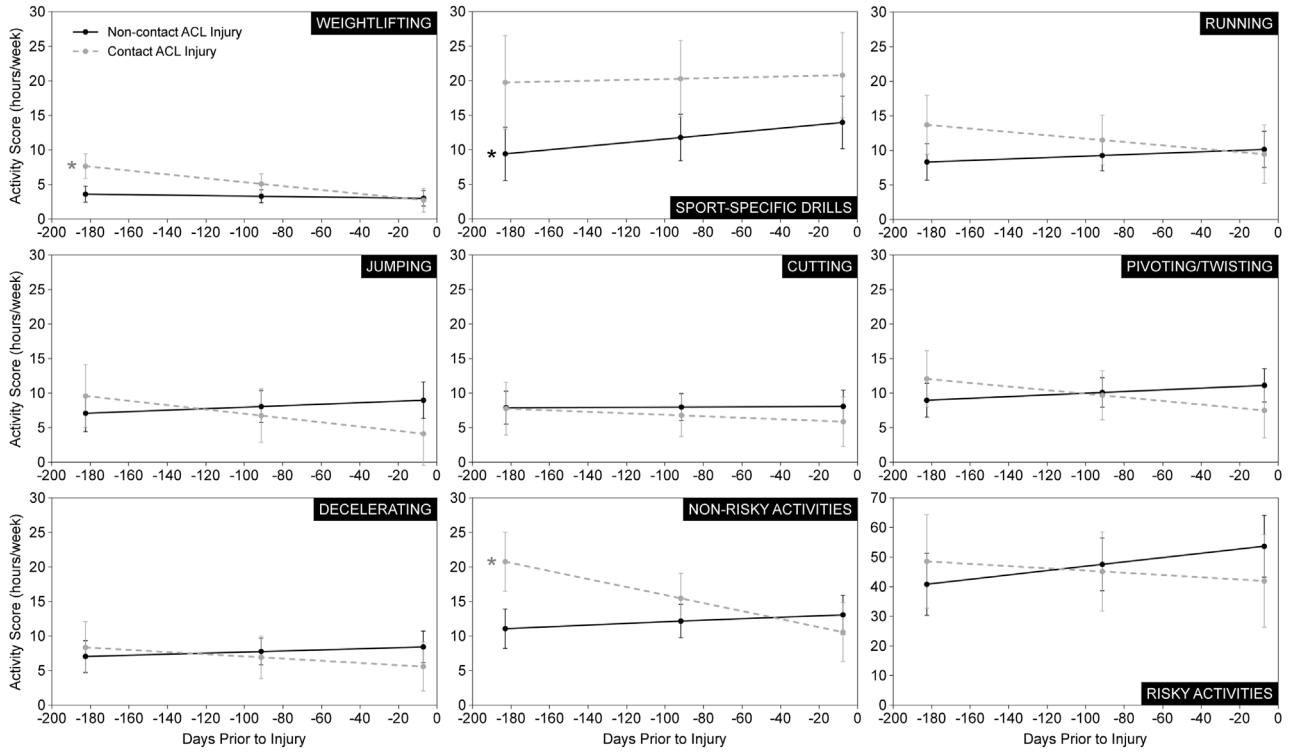

FIGURE 2
Estimated marginal mean activity scores. Estimated marginal mean activity scores (± 1 standard error) at each time point (6 months, 3 months, 7 days) prior to injury for each ACL injury group. Data are presented for all seven athletic activities and maneuvers in addition to the summative risky and non-risky activities. *Statistically significant slope ($p < 0.10$) indicating a significant change in athletic activity and maneuver levels leading up to ACL injury.

88.2%/M: 100%), 86.4% (F: 87.9%/M: 81.8%), 76.6% (F: 77.1%/M: 75.0%), 95.6% (F: 94.1%/M: 100%), and 82.2% (F: 79.4%/M: 90.9%) participated in weightlifting, sport-specific drills, running, jumping, cutting, pivoting/twisting, and decelerating activities, respectively.

The statistics of the linear mixed-effects regression models for all athletic activities or maneuvers of all patients are presented in Table 2. A graphical representation of the change in each activity or maneuver's score over the 6-month period leading up to ACL injury, for each injury group, can be found in Figure 2. Interestingly, patients who sustained a non-contact ACL injury significantly increased their sport-specific drills activity (i.e., score) in the 6-month period leading up to injury; meanwhile no change was observed in the sport-specific drills activity in the patients who sustained a contact injury (Table 2; Figure 2). In addition, the linear mixed-effects regression model for weightlifting revealed that patients engaged in significantly decreasing weightlifting activity over the 6-month period leading up to injury (Table 2; Figure 2). When breaking down the results by injury type, contact ACL injury patients significantly decreased their weightlifting activity leading up to injury; meanwhile no change was noted in weightlifting activity in the non-contact injury patients (Table 2; Figure 2). Similarly, contact ACL-injured patients also significantly decreased their combined non-risky activities or maneuvers, which includes weightlifting, leading up to injury (Table 2; Figure 2).

The change in the frequency/intensity of the running, jumping, cutting, pivoting/twisting, and decelerating activities or maneuvers, as well as the summative risky and non-risky activities or maneuvers in the 6 months leading up to ACL injury exhibited an interesting trend. Levels tended to increase in the non-contact ACL injury participants but tended to decrease in the contact injury participants (Figure 2). In fact, the change leading up to injury in levels of weightlifting, jumping, pivoting/twisting, and the summative non-risky activities/maneuvers was significantly different between patients who sustained a contact ACL injury and those who sustained a non-contact ACL injury (Table 2; Figure 2).

4 Discussion

This retrospective study investigated whether the frequency and/or intensity at which ACL-injured patients performed various athletic activities or maneuvers significantly changed in the 6-month period leading up to their ACL injury. Our results indicate that indeed there was a change in the frequency/intensity of several activities or maneuvers in the months leading up to injury. There was a significant increase in the frequency/intensity of sport-specific drills in patients who sustained a non-contact ACL injury; and there was a significant decrease in the frequency/intensity of weightlifting and summative non-risky activities or maneuvers in those who

sustained a contact ACL injury. In general, this study illustrated a trend for levels of athletic activities or maneuvers deemed "risky" in terms of ACL injury to increase leading up to a non-contact ACL injury and for levels of 'non-risky' activities/maneuvers to decrease leading up to a contact injury. The rate at which the levels of various athletic activities or maneuvers changed in the 6-month period leading up to injury was significantly different between non-contact and contact ACL injury patients. In particular, this rate of change was significantly different between injury mechanism groups for jumping, pivoting/twisting, and weightlifting.

The significant increase in the frequency and/or intensity of sport-specific drills activity leading up to a non-contact ACL injury is consistent with the notion that the ACL can fail due to the accumulation of microdamage when subjected to submaximal repetitive loading (i.e., 'tissue fatigue' injury). The general trend of increasing levels of athletic activities and maneuvers deemed 'risky' in terms of ACL injury risk amongst non-contact ACL injury patients prior to injury is also in line with this notion. This is because a marked increase in the intensity and/or frequency of ACL-straining athletic activities and maneuvers, without allowing enough time for the ACL to adapt, may cause the ligament to transition from a state of homeostasis where its repair rate matches its rate of microdamage to a catabolic state, leading to progressive weakening and then the eventual failure of the ACL under normal conditions. A good example of this tissue homeostasis, or lack thereof, in the sports world is the UCL injury in baseball pitchers. It has been established that the UCL can sustain a "tissue fatigue" injury whereby microdamage accumulates in the ligament of pitchers who repetitively throw at high velocities without scheduling enough rest to allow for adequate ligament repair (Mirowitz and London, 1992). This careful balancing act between catabolic and anabolic processes exists in other tissues. For example, in skeletal muscle and bone remodeling, there is a coordinated process that balances protein degradation vs. protein synthesis and bone resorption vs. bone formation, respectively (McCarthy and Esser, 2010; Raggatt and Partridge, 2010). As for the ACL, evidence of such a "tissue fatigue" injury mechanism comes from *ex vivo* ACL tissue, retrieved from non-contact ACL injured patients during their reconstruction surgery, when compared to ACL tissue from repeatedly-loaded cadaveric knees (Chen et al., 2019; Kim et al., 2022). These cadaveric knees had undergone repeated loading that simulated pivot-landings (i.e., knee compression, flexion, and internal tibial rotation) and were investigated for damage to the ACL. Interestingly, the cadaveric ligaments exhibited disruptions of the collagen in the form of unraveling and voids following repetitive loading (Kim et al., 2022), the same pattern of structural damage that was found in the ACL explants from non-contact ACL injury patients undergoing reconstruction (Chen et al., 2019). In short, results of the present study represent the first *in vivo* evidence that significant increases in sports training dosage, in particular ACL-straining athletic activities and maneuvers, are associated with non-contact ACL injury. This provides indirect support for the "tissue fatigue" injury mechanism for which the ACL does not have enough time to fully repair, thereby leading to the accumulation of microdamage, weakening of the ligament and eventual failure.

Demonstrating that the ACL can fail via a "tissue fatigue" injury mechanism, whereby a rapid increase in the intensity of ACL-straining athletic activities and maneuvers can affect the balance between ACL degradation and recovery, has several implications for injury prevention. For instance, monitoring and limiting athletes' levels of ACL-straining activities and maneuvers, similar to the use of the pitch count for baseball players in UCL injury prevention, may help prevent some non-contact ACL injuries (Wojtys et al., 2016). Doing so would give the ACL enough time to recover between bouts of large strain and thus ligament homeostasis could be maintained. This does not preclude training activities and maneuvers known to not significantly load the ACL such as running without sharp turns, for example,. For ACL 'tissue fatigue' injury prevention efforts to be successful, many factors that are currently poorly characterized need to be investigated and taken into consideration. First, the healing rate of the ACL is poorly understood although we know that it is relatively slow compared to other tissues such as bone, cartilage, and muscle (Panjabi, 2006; Jung et al., 2009; Nyland et al., 2022), thus making the ACL particularly susceptible to the accumulation of microdamage from repetitive submaximal loading cycles. Second, the location of the microdamage within the ligament should also be considered because the ligament insertion sites take longer to remodel than other regions (Arnoczky, 1983; Toy et al., 1995). Third, sex differences in ACL healing rate should be investigated. For example, female athletes experience ACL injuries 2–10 times more frequently than males due to several established physiological and anatomical factors (Sturnick et al., 2015; Shultz et al., 2019; Barnum et al., 2021). Perhaps a sex difference in ligament healing rate is another contributing factor. If such a sex difference exists, it could have affected our results since the majority of the ACL-injured patients were females. Fourth, nutrition and sleep should be considered as well given that these factors can contribute to the integrity of the ACL's extracellular matrix (Dodt et al., 1997; Kjaer et al., 2009; Nyland et al., 2022). Lastly, age differences exist in the turnover rate of collagen and should be taken into consideration (Sivan et al., 2008). Additionally, with the hamstrings and quadriceps known as agonists and antagonists, respectively, to the ACL, athletes can train to perform their respective sport maneuvers in a way that minimizes ACL strain but without compromising performance (Withrow et al., 2006; 2008; Wojtys et al., 2016). This concept of optimizing maneuvers is important given the known inverse relationship between ACL load and number of loading cycles to ligament failure (Lipps et al., 2013). *In vitro* repetitive loading of human knees revealed that when the magnitude of the repetitive load applied to the knee decreased, the number of loading cycles needed to fail the ACL increased. Therefore, if athletes are able to optimize their maneuvers to consistently keep the strain placed on the ACL below a certain threshold by using a monitoring system, for example, it is possible that many ACL fatigue injuries could be prevented effectively.

The significant decrease in weightlifting and summative non-risky athletic activities and maneuvers in contact ACL-injured patients is an interesting result. Given that the summative non-risky activity score combined the scores of only the weightlifting and running activities, this significant decrease in non-risky activity levels was most likely due to the significant decrease in weightlifting activity level. Accordingly, this decrease in weightlifting activity might have resulted in the patients' muscle strength not being maintained during the 6-month period leading up to injury. Although most contact injuries are probably a result of a single event that places the healthy ACL under excessively large strain, where the accumulation of microdamage in the ligament plays a minimal role, a decrease in muscle strength may cause the knee

to become more susceptible to injury for a variety of reasons. Muscle strength is important for controlling lower limb dynamic stability and aids in resisting muscle fatigue, which is associated with decreased performance (Schipplein and Andriacchi, 1991; Besier et al., 2003; Hunter et al., 2004). In fact, loss of lower extremity muscle strength has been associated with greater risk of sustaining a traumatic knee injury in females (Ryman Augustsson and Ageberg, 2017). Therefore, the reduction in weightlifting activity may have placed the knee and its ACL at an increased risk of injury.

Some limitations of our study should be noted. First, since this was a retrospective survey, it is subject to recall bias. We relied on the patients to accurately recall their activity levels up to 6 months before their ACL injury. It probably helped that the majority (57%) of the patients completed the questionnaire within 1 month of injury, with 85% completing it within 2 months, and 89% within 3 months, thereby limiting the recall period and potentially reducing recall bias. In addition, many athletes have set schedules for their extracurricular activities, which can ease recall. Second, the questionnaire only assessed levels of athletic activities and maneuvers at three time points (6 months, 3 months, and 1 week) prior to injury. This was implemented by design in order to balance the collection of detailed data with the questionnaire's time burden on the patients and their willingness and ability to complete it adequately. Thus, it did not capture any changes in activities or maneuvers that may have occurred during the 3-month intervals between these time points. Nevertheless, we are confident that this 3-month interval captured any significant changes that would affect ACL integrity given the slow turnover rate of the ACL collagen (Rucklidge et al., 1992). Third, our sample size was modest, especially when assessing non-contact and contact ACL injury groups separately; nevertheless, we found statistically significant changes in activity levels in the 6-month period leading up to ACL injury. Building on our finding that sport-specific drills activity significantly increased in non-contact injury patients, future research should increase the sample size with a focus on non-contact injury patients. This could uncover other athletic activities or maneuvers that exhibit a similar pattern. Fourth, there were missing data for several patients. However, our statistical approach—the linear mixed-effects regression model—allowed for the retention of all data from all patients as opposed to alternate approaches (e.g., analysis of variance), which removes cases with missing data from the analysis. Lastly, we did not account for other contributing factors to injury risk such as the patients' nutrition, sleep, and use of a knee brace. However, given that the athletes were largely amateurs (school-aged or recreational adult athletes), we suspect that such factors confounded our study's results minimally, if at all.

Ultimately, the levels of several athletic activities/maneuvers of ACL-injured patients changed in the 6 months leading up to their injury, including a significant increase in levels of sport-specific drills among patients who sustained a non-contact injury and a significant decrease in weightlifting activity among those who sustained a contact injury. These results provide evidence that a rapid increase in levels of activities/maneuvers known to apply large forces on the ACL may be a risk factor for injuries to this ligament, especially for non-contact ACL injuries, which has implications for training and injury prevention. Future research should aim to provide direct *in vivo* evidence of microdamage accumulation in the ACL, for instance by means of serum biomarkers (Svoboda et al., 2016) or magnetic resonance imaging (Barnes et al., 2023).

Data availability statement

The raw data supporting the conclusion of this article will be made available by the authors, without undue reservation.

Ethics statement

The studies involving human participants were reviewed and approved by Institutional Review Board at the University of Michigan. Written informed consent to participate in this study was provided by the participants and/or their parent/legal guardian.

Author contributions

JA-M and EW contributed to the conception of the study. MB, JA-M, and EW contributed to the study design. LG and EW contributed to data collection. LG, MB, JA-M, and EW contributed to data analysis. MB contributed to statistical analysis. LG, MB, JA, and EW contributed to manuscript development.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Supplementary material

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fphys.2023.1166980/full#supplementary-material

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The effect of blood flow restriction training combined with electrical muscle stimulation on neuromuscular adaptation: a randomized controlled trial

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Objective: Low-intensity resistance training (≤25% 1RM) combined with blood flow restriction training (BFRT) is beneficial to increasing muscle mass and muscle strength, but it cannot produce increased muscle activation and neuromuscular adaptation, as traditional high-intensity strength training does. The purpose of this study is to investigate the effects of independently applying BFRT and electrical muscle stimulation (EMS), as well as combining the two methods, on muscle function.

Methods: Forty healthy participants with irregular exercise experiences were randomly assigned to four groups: BFRT-alone group (BFRT, n = 10), EMS-alone group (EMS, n = 10), BFRT combined with EMS group (CMB, n = 10), and the control group (CTR, n = 10). All participants received low-intensity squat training at a load of 25% 1RM 5 times/week for 6 weeks. Cross-sectional area (CSA) and electromyographic root mean square (RMS) in the rectus femoris, as well as peak torque (PT) of the knee extensor, were measured before and following a 6-week intervention.

Results: Following the 6-week intervention, the increases in muscle activation in the CMB group were statistically higher than those in the BFRT group ($p < 0.001$), but not different from those in the EMS group ($p = 0.986$).

Conclusion: These data suggest that the combination of BFRT and EMS for low-intensity squat training improved the muscle strength of the lower limbs by promoting muscle hypertrophy and improving muscle activation, likely because such a combination compensates for the limitations and deficiencies of the two intervention methods when applied alone.

KEYWORDS

blood flow restriction, electromyostimulation, muscle activation, muscle strength, muscle hypertrophy

1 Introduction

Loss of muscle mass in the lower extremities and reduced muscle strength due to decreased neuromuscular adaptation are important causes of serious health problems such as falls, disability, and even death (Visser et al., 2005). Squat training is an effective method to prevent muscle atrophy and improve muscle strength in the lower extremities (Kraemer et al., 2017). According to the American College of Sports Medicine (ACSM), resistance strength training at a load of ≥70% 1-repetition maximum (1RM) can effectively induce muscle hypertrophy and improve muscle strength (Kraemer et al., 2009). However, for postoperative rehabilitation patients, athletes with skeletal muscle injuries, the elderly, and other populations with special needs, high-load, and high-intensity resistance training is not practical. In addition, for ordinary people who have no prior resistance training experience, such training may increase the risk of sports injury (Gardiner et al., 2020).

Blood flow restriction training (BFRT) is a method of applying equipment with special material to exert pressure on the proximal end of the limbs during exercise to enhance the training effect. Numerous studies have confirmed that low-intensity resistance training (≤25% 1RM) combined with BFRT is beneficial for increasing muscle mass and muscle strength (Colomer-Poveda et al., 2017; Korakakis et al., 2018; Ladlow et al., 2018). Therefore, BFRT can be used as a recommended method for people who are unable to perform traditional high-load resistance strength training. However, BFRT has an inherent limitation. The underlying mechanism of BFRT-based strength training is dominated by muscle hypertrophy, as it cannot produce increased muscle activation and neuromuscular adaptation, as traditional high-intensity strength training does (Manini and Clark, 2009; Cook et al., 2013).

Electrical muscle stimulation (EMS) uses a low-frequency rectangular pulse electric current to stimulate a specific muscle, resulting in involuntary muscular contraction. Numerous studies have shown that EMS is beneficial for improving muscle strength (Šarabon et al., 2020; Kemmler et al., 2021; Nishikawa et al., 2021). However, there is no consensus on the existence of an "optimal" frequency range. Filipovic et al. stated that frequencies of approximately 76 Hz lead to optimal strength development (Filipovic et al., 2011). Previous studies suggested that the muscle strengthening seen following EMS results from a reversal of voluntary recruitment order with a selective augmentation of type-II muscle fibers (Gregory and Bickel, 2005; Jubeau et al., 2007; Sheffler and Chae, 2007). Because type-II fibers have a higher specific force than type-I fibers, selective augmentation of type-II muscle fibers will increase the overall strength of the muscle. However, the main limitation of EMS for inducing muscle hypertrophy and improving muscle strength is directly related to the stimulation intensity. The intensity required to effectively increase strength and hypertrophy is particularly prone to induce prohibitive pain and rapid muscle fatigue (Filipovic et al., 2011), thus reducing the participants' compliance. Recent studies found that the combination of BFRT and low-intensity EMS might avoid the discomfort caused by high current intensity when EMS has been applied alone (Natsume et al., 2015; Slysz and Burr, 2018; Slysz et al., 2021).

Therefore, the purpose of this study was to investigate the effects of BFRT alone, EMS alone, and BFRT + EMS on muscle volume, muscle activation, and muscle strength. Low-intensity squat training (≤25% 1RM) was used as a control. By comparing the changes in the parameters related to lower limb muscle volume, muscle activation, and muscle strength of subjects under different intervention methods, the effects of each intervention method were confirmed and the related mechanisms of their influence on muscles were clarified, to provide a research basis for the promotion and application of combined squat training in clinical rehabilitation. We hypothesize that the combination of BFRT and EMS will lead to a greater training effect than either of these interventions in isolation.

2 Materials and methods

2.1 Participants

A total of forty healthy young men were recruited from the Chengdu Sport University. The inclusion criteria were as follows: 1) males, 2) good physical health (screening exercise risk through the PAR-Q), 3) low levels of physical activity (scored by the IPAQ), 4) naive to squat training, BFRT training, or EMS training and 5) absence of lower limb injury. After being advised of the purpose and potential risks of the study, all participants provided written informed consent, and the study was approved by the ethics subcommittee of Chengdu Sport University (CDSUEC_2020_17_57) and conducted in accordance with the Declaration of Helsinki.

2.2 Study design

This was a randomized study. Random sequences generated by the SPSS version (SPSS Inc., United States) were hidden in consecutively numbered (1–40), sealed, opaque envelopes, where the allocation ratio was 1:1:1:1. Participants were randomly allocated to either the BFRT-alone group (BFRT, n = 10), the EMS-alone group (EMS, n = 10), the BFRT combined with EMS group (CMB, n = 10), or the control group (CTR, n = 10). When the researchers confirmed that the participants met the inclusion/exclusion criteria, the envelopes were opened in sequence and the participants were assigned to the corresponding groups. All outcomes were measured by rehabilitation therapists with standardized training and professional technical qualifications, and each participant was measured by the same evaluator before and after the intervention to avoid errors.

2.3 Intervention

Each participant underwent an intervention protocol consisting of 30 19-min sessions, which took place once per day, 5 days a week, and lasted for 6 weeks (Figure 1). Each session was supervised by the same researcher. All experiments were carried out at the Institute of Sports Medicine and Health of Chengdu Sport University.

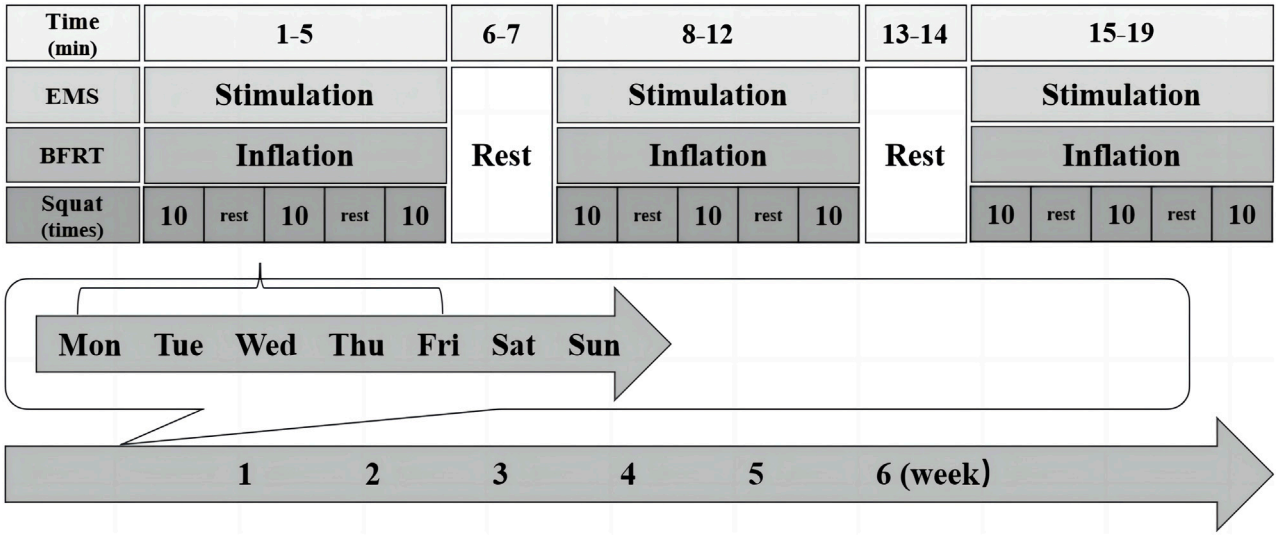

FIGURE 1
The timeline of the entire experimental intervention. All participants underwent exercise intervention for 19 min/session, once/day, 5 days/week, and lasted for 6 weeks.

FIGURE 2
The BFRT group. A 5-cm wide tourniquet cuff was placed at the groin crease, and according to the leg circumference inflated to the appropriate pressure (≤50 cm, 200 mmHg; 51–55 cm, 250 mmHg; 56–59 cm, 300 mmHg; ≥60 cm, 350 mmHg). Each session comprised 3 cycles of 5 min of inflating and 2 min of deflating.

2.3.1 CTR group

High-bar squat training (Murawa et al., 2020) at 25% 1RM was performed in the first, third, and fifth minutes of the session, and the rest was performed in the second and fourth minutes. The rhythm of the squat was controlled by a metronome 10 times/min (3 s concentric/3 s eccentric). Squat 30 reps per set, with a 2-min interval, and repeat for 3 sets. At the end of the second and fourth weeks of intervention, 1RM was reassessed periodically to appropriately adjust for resistance.

2.3.2 BFRT group

BFRT was performed with an automated tourniquet system (B-strong, United States). The squat program was the same as that of the CTR group. Meanwhile, a 5-cm wide tourniquet cuff was placed at the groin crease, and according to the leg circumference inflated to the appropriate pressure (≤50 cm, 200 mmHg; 51–55 cm, 250 mmHg; 56–59 cm, 300 mmHg; ≥60 cm, 350 mmHg) (Jia Wei et al., 2019). Each session comprised 3 cycles of 5 min of inflating and 2 min of deflating. As shown in Figure 2.

2.3.3 EMS group

EMS was performed with an electrical muscle stimulation instrument (Twin Stim Plus: Tens Unit 7,000, United States). The squat program was the same as that of the CTR group. The EMS protocol consisted of a duty cycle of 3 s ON (stimulation) and 3 s OFF (no stimulation) with the pulse width set at 400 μs, the intensity set at 50 mA, and the stimulation frequency set at 75 Hz (Filipovic et al., 2011; Seyri and Maffiuletti, 2011). Before performing the squat, an exercise pointer was used to accurately locate the most obvious point of muscle contraction. Then, two negative electrodes (5 × 5 cm) were placed near

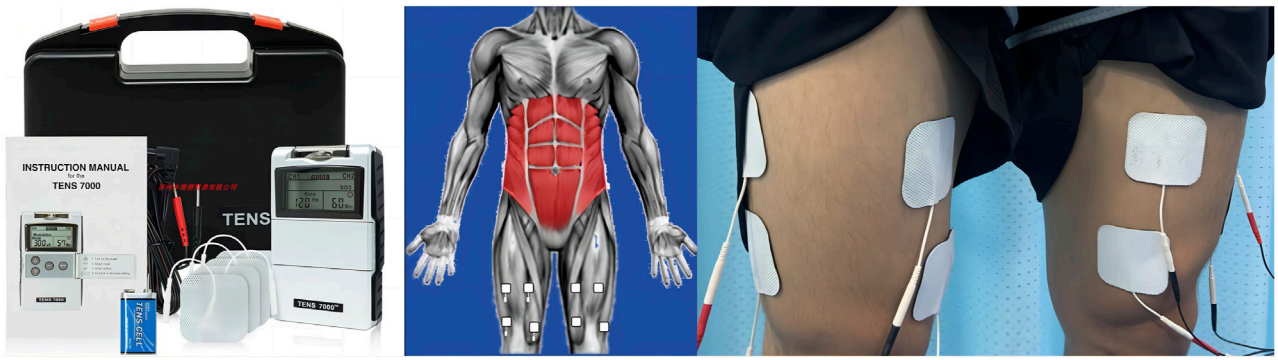

FIGURE 3
The EMS group. Two negative electrodes (5 × 5 cm) were placed near the groin crease approximately 10 cm below, and the other two positive electrodes (5 × 5 cm) were placed as close as possible to the movement point of the vastus lateralis and medial muscles. The EMS protocol consisted of a duty cycle of 3 s ON (stimulation) and 3 s OFF (no stimulation) with the pulse width set at 400 μs, the intensity set at 50 mA, and the stimulation frequency set at 75 Hz. Each session comprised 3 cycles of 5 min of stimulation and 2 min of rest.

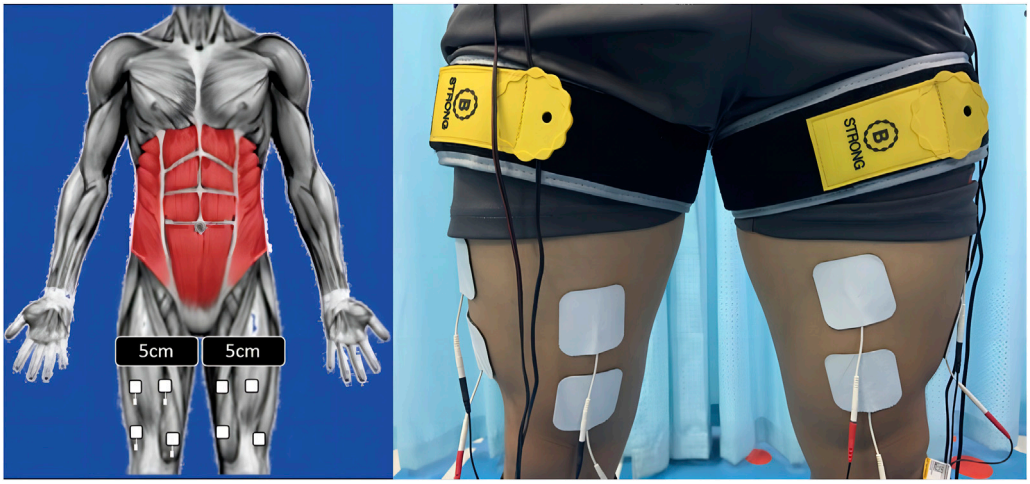

FIGURE 4
The CMB group. In the CMB group, BFRT and EMS were carried out simultaneously in the same way as above. The squat program was the same as that of the CTR group.

the groin crease approximately 10 cm below, and the other two positive electrodes (5 × 5 cm) were placed as close as possible to the movement point of the vastus lateralis and medial muscles. Each session comprised 3 cycles of 5 min of stimulation and 2 min of rest. As shown in Figure 3.

2.3.4 CMB group

BFRT and EMS were carried out simultaneously in the same way as above. The squat program was the same as that of the CTR group. As shown in Figure 4.

2.4 Outcomes

2.4.1 Muscle strength and muscle activation

The muscle activation test was performed at the same time as the isokinetic muscle strength test. Before the test, participants underwent a 10-min warm-up session. A wireless surface electromyography (sEMG) system (Myon 320, Schwarzenberg, Switzerland) was used to collect electrical signals from muscles during exercise. Electrodes (Ag/AgCl) were affixed to the rectus femoris on both sides of the participant, and the electrodes were 2-3 cm apart. Then, the participants were asked to sit relaxed, and the baseline signals and signal quality of all electrode signals were detected through knee flexion and extension. Then, all participants underwent strength testing of the knee extensors using an isokinetic dynamometer (Con-Trex MJ, Switzerland). Knee extension maximal voluntary contractions were measured at 90° with participants' seats at 90° hip flexion. Each leg was positioned on the dynamometer by aligning the participants' knee with the axis of the lever arm and attaching the dynamometer arm approximately 3 cm proximal to the medial malleolus. Participants were instructed to contract as "fast and

forcefully as possible". Standard verbal encouragement and visual feedback were provided during each contraction. Before the formal test, participants tried 3 moderate strength exercises to familiarize themselves with the procedure. The test speed was set at 30°/s, which was repeated for 5 contractions.

The outcomes of muscle strength were selected as the peak torque (PT) of knee extension. The sEMG signal was analyzed in the time domain, and the selected index was the root mean square (RMS). sEMG preprocessing: The raw sEMG sampling frequency was 2000 Hz, and the sEMG data were preprocessed by low-pass and high-pass filtering (frequency 10–400 Hz) and full-wave rectification. To avoid the influence of different participants' RMS before and after the experiment, such as the position of electrode paste and cortical thickness, the maximum normalization method was used to standardize the RMS values.

2.4.2 Muscle thickness

The cross-sectional area (CSA) of the rectus femoris was assessed via ultrasound (PHILIPS, Netherlands) as an index of the whole rectus femoris muscle volume. The measurement position was supine. The site was landmarked to ensure consistency of measures (from the midpoint of the patella to the middle of the 30%, 50%, and 70% cross-sections of the greater trochanter of the femur). The average value of the three locations was taken as the final result. Using B-mode ultrasound with a linear-array probe (L12-3, 3–12 MHz, Philips) with minimal pressure was applied to avoid compression of the underlying tissue. Images were analyzed immediately post-acquisition, and muscle thickness was assessed by measuring the distance (cm) from the superficial to the deep aponeurosis at the exact center of the fasciae (QLAB Software, PHILIPS). The final CSA of the rectus femoris is the average of the left and right sides.

2.5 Statistical analysis

The descriptive statistics are reported as the mean ± standard deviation ($\bar{x} \pm SD$). The difference (Delta) between the pre- and post-intervention muscle volume, muscle activation, and muscle strength were calculated between week 0 and week 6. For each outcome measure, a one-way ANOVA was performed. The dependent variable was the Delta of the specific outcome measure, and the factor was the Intervention Methods, which had four levels: CTR, BFRT, EMS, and CMB. In case of significant F, a Bonferroni *post hoc* analysis will be performed to assess the difference across the four intervention methods. Statistical significance was set at $p < 0.05$, and all analyses were performed with SPSS version 25.0 (SPSS Inc., United States).

3 Results

3.1 Descriptive characteristics

The descriptive characteristics of 40 participants who completed the entirety of the 6-week intervention are provided in Table 1. Before the intervention, no significant differences were observed between the four groups in terms of age (p = 0.736), height (p = 0.589), weight (p = 0.604), and body mass index (BMI, p = 0.490).

TABLE 1 Descriptive characteristics of the participants.

Items	CTR (n = 10)	EMS (n = 10)	BFRT (n = 10)	CMB (n = 10)	p-value
Age (years)	20.9 ± 1.1	20.8 ± 1.5	20.4 ± 1.7	20.5 ± 1.0	0.736
Height (cm)	172.4 ± 8.1	171.6 ± 6.9	172.7 ± 6.4	170.6 ± 9.4	0.589
Weight (kg)	67.6 ± 10.2	67.2 ± 9.7	65.3 ± 11.2	66.1 ± 9.1	0.604
BMI (kg/m^2)	20.9 ± 2.1	21.3 ± 2.4	21.1 ± 2.7	20.7 ± 3.0	0.490

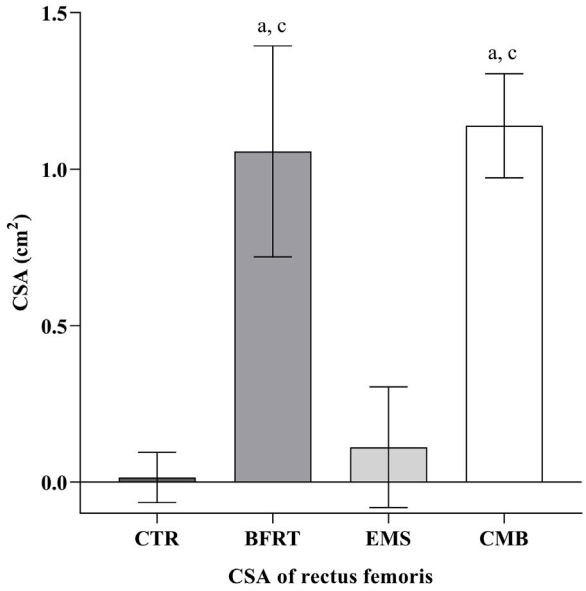

FIGURE 5
The CSA of rectus femoris in each group. Letters a, b, and c represent comparison with CTR group, BFRT group, and EMS group, respectively. Difference from CTR, [a] $p < 0.05$; Difference from BFRT, [b] $p < 0.05$; Difference from EMS, [c] $p < 0.05$.

3.2 Muscle volume

Following the 6-week intervention, the changes in CSA, separated by all groups, are shown in Figure 5. There was a greater CSA increase in the CMB group compared to the EMS group ($p < 0.05$) and the CTR group ($p < 0.05$), but was not different from that in the BFRT group (p = 0.916). It also confirmed a greater CSA increase in the BFRT group than in the CTR group ($p < 0.05$) and the EMS group ($p < 0.05$). However, there was no significant change between the CTR group and the EMS group (p = 0.262).

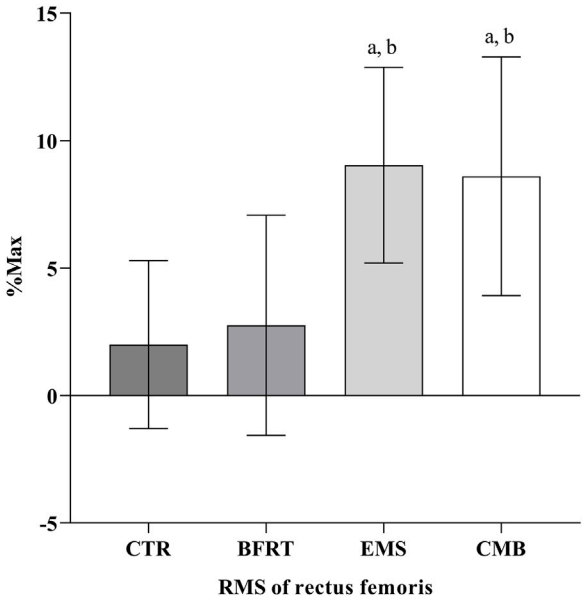

FIGURE 6
The RMS of rectus femoris in each group. Letters a, b, and c represent comparison with CTR group, BFRT group, and EMS group, respectively. Difference from CTR, [a] $p < 0.05$, [aa] $p < 0.01$; Difference from BFRT, [b] $p < 0.05$; Difference from EMS, [c] $p < 0.05$.

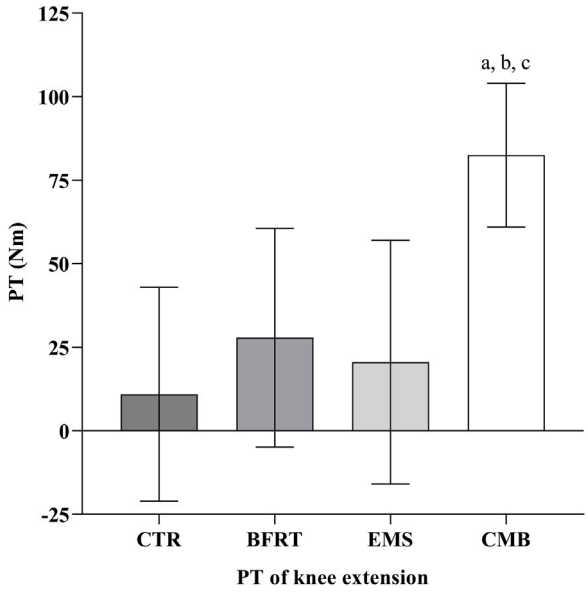

FIGURE 7
The PT of knee extensor in each group. Letters a, b, and c represent comparison with CTR group, BFRT group, and EMS group, respectively. Difference from CTR, [a] $p < 0.05$, [aa] $p < 0.01$; Difference from BFRT, [b] $p < 0.05$; Difference from EMS, [c] $p < 0.05$.

3.3 Muscle activation

Following the 6-week intervention, the changes in RMS, separated by all groups, are shown in Figure 6. There was a greater RMS increase in the CMB group compared to the BFRT group ($p < 0.05$) and the CTR group ($p < 0.05$), but was not different from that in the EMS group ($p = 0.986$). It also confirmed a greater RMS increase in the EMS group than in the CTR group ($p < 0.05$) and the BFRT group ($p < 0.05$). However, there was no significant change between the CTR group and the BFRT group ($p = 0.934$).

3.4 Muscle strength

Following the 6-week intervention, the changes in PT, separated by all groups, are shown in Figure 7. There was a greater PT increase in the CMB group compared to the CTR group ($p < 0.05$), the EMS group ($p < 0.05$), and the BFRT group ($p < 0.05$). However, there was no significant change in the BFRT group compared to the CTR group ($p = 0.321$), and the EMS group ($p = 0.879$). And also, there was no significant change in the EMS group compared to the CTR group ($p = 0.763$).

4 Discussion

Considering the important role of lower extremity muscle strength in daily activities and sports, as well as a more suitable exercise mode for clinical skeletal muscle rehabilitation patients, elderly individuals with muscular atrophy, and other special populations, this study chose low-intensity (25% 1RM) squat training as the exercise intervention method. The applications of this method could have important implications for sports rehabilitation.

Studies on the process of muscle strength growth have shown that early strength growth (0–10 weeks) in the early phase is due to increased neuromuscular adaptation, while strength growth in the later period (after 10 weeks) is related to muscle hypertrophy (Kraemer et al., 1996). In this study, BFRT combined with low-intensity squat training was used to induce significant muscle hypertrophy after 6 weeks, proving that BFRT combined with low-intensity resistance training seems to produce a muscle hypertrophy effect similar to that of the traditional high-intensity resistance exercise, which is consistent with the results of previous studies. At present, the research on the effect of EMS is mostly focused on muscle strength (de Oliveira et al., 2021; Nishikawa et al., 2021; Rodrigues-Santana et al., 2021), and there are few studies on the effect of muscle volume. In this study, it was found that 6 weeks of EMS combined with low-intensity squat training could promote the volume of RF, but the improvement effect was significantly less than that of the BFRT group and CMB group. This may be related to the intensity of electrical stimulation and the intervention cycle selected in this experiment. To avoid the discomfort caused by electric current to the participants, the minimum threshold value (75 Hz), which can effectively stimulate the increase in muscle strength, was selected in this study. In terms of the combination

of BFRT and EMS, Gorgey et al. found that the acute effects of EMS + BFRT may suggest that an increase in flow-mediated dilation may partially contribute to skeletal muscle hypertrophy, and BFRT combined with EMS could be a strategy to increase skeletal muscle size (Gorgey et al., 2016). Natsume et al. found in animal experiments that BFRT combined with EMS can induce muscle hypertrophy more than BFRT alone, resulting in increased protein expression levels related to muscle hypertrophy and accumulation of metabolites, as well as promoting muscle fatigue and increasing gastrocnemius muscle mass (Natsume et al., 2019). The results of this study showed that the combined BFRT and EMS could induce muscle volume changes more than EMS alone, which was consistent with previous studies, but the specific mechanism needs to be further studied.

Increased recruitment of muscle fibers is thought to be one of the underlying neural mechanisms that cause muscle strength to increase (Loenneke et al., 2011). However, our study found that combined BFRT and EMS were significantly better than BFRT alone in promoting muscle activation, and BFRT combined with low-intensity squat training did not seem to produce the same increase in muscle activation as the traditional high-intensity training (Manini and Clark, 2009; Cook et al., 2013). The main factor by which BFRT combined with low-intensity squat training improves muscle strength may be the muscle hypertrophy effect, while neuromuscular adaptation has little effect. Yamanaka et al. showed that compared with low-intensity BFRT (30% 1RM), the training effect of high-intensity training (75% 1RM) and combined training (2 days low-intensity BFRT+1 day high-intensity training) was mainly due to neural adaptation (Yamanaka et al., 2012). Therefore, it was suggested that BFRT can be used in combination with traditional high-intensity training to produce more effective stimulation of muscle volume and neuroadaptation. Previous studies have found that BFRT can cause nerve numbness and other adverse reactions (Cook et al., 2013). Therefore, we believe that the failure of BFRT to enhance neuromuscular adaptability may be related to the limitation of nerve conduction while restricting blood flow. At present, no study has explicitly investigated the effect of BFRT combined with EMS on the electrical activity of muscles. The results of this study found that compared with low-intensity training + BFRT, low-intensity training combined with BFRT + EMS had a better effect on the increase in muscle activation. The combination of BFRT and EMS can simultaneously achieve effective improvement of muscle volume and neuromuscular adaptation. Training combined with EMS can compensate for the weakness of BFRT limiting the increase in muscle fiber recruitment, and low-intensity training is safer than high-intensity training. In addition, studies have shown that after BFRT, the concentration of inorganic phosphate in muscle increases significantly, resulting in more obvious peripheral fatigue (Head et al., 2020). After BFRT, inorganic phosphate in type I and type II muscle fibers was rapidly consumed, and EMS increased muscle fiber recruitment to maintain power output. This may also be a possible mechanism for increased muscle activation in the CMB group.

The role of BFRT in promoting muscle strength has been confirmed by a large number of studies (Shinohara et al., 1998; Takarada et al., 2000). The results confirmed that BFRT combined with low-intensity squat training can improve the muscle strength of participants who exercise irregularly, and it was believed that the main reason for the improvement of the lower limb muscle strength caused by BFRT combined with low-intensity squat training may be related to the significant increase in muscle volume.

The improvement of muscle strength depends not only on the increase in muscle volume but also on the increase in motor unit recruitment. EMS can improve the innervation of the central system and coordinate the agonistic and antagonistic muscles, thus promoting the recruitment of motor units, activating more muscle fibers, and enhancing muscle strength (Kemmler et al., 2018). The mechanism by which EMS promotes muscle strength is closely related to the intensity and frequency of stimulation. Cunha et al. found that 8 weeks of EMS training (70 Hz) can effectively improve volleyball players' lower limb muscle strength and explosive power (da Cunha et al., 2020). Toth et al. pointed out that EMS (400 μs, 50 Hz) can prevent quadriceps atrophy after ACL surgery and improve the maximum muscle strength of patients (Toth et al., 2020). Pantović et al. showed that EMS and traditional high-intensity resistance strength training had the same training effect in improving muscle strength (Pantović et al., 2015). Our study found that 6 weeks of EMS combined with low-intensity squat training effectively improved lower limb muscle strength, and the intervention effect was similar to that of BFRT combined with low-intensity squat training, which may be related to more significant muscle activation in the EMS group. In terms of the combination of BFRT and EMS, Gorgey et al. found that the combination of BFRT and EMS can effectively improve the wrist strength of patients with incomplete quadriplegia (Gorgey et al., 2016). Natsume et al. showed that the combination of BFRT and EMS can significantly increase the maximum knee extension strength of young men with irregular exercise habits (Natsume et al., 2019). Slysz et al. found that 6 weeks of EMS combined with BFRT training effectively improved the leg lift strength of lower limbs (Slysz and Burr, 2018). The above scholars all believe that the increase in muscle strength is mainly related to muscle hypertrophy. However, the results of this study showed that when BFRT was used, whether or not EMS was used together had a significant influence on muscle strength. Similarly, when using EMS, whether BFRT was combined with EMS also had a significant influence on muscle strength. This suggests that the increase in muscle strength caused by BFRT and EMS is related to neuromuscular adaptation and muscle hypertrophy.

The current study has some limitations. First, the sample of this study was restricted to young, healthy men. Thus, we acknowledge that our findings may not apply to other populations. In future studies, patients with musculoskeletal diseases can be recruited as research participants to further verify the clinical application value. Second, due to a large number of measurements and participants involved in this experiment, all measurements were only collected before and after the intervention. Therefore, future studies can appropriately focus on certain major measurements and exclude individual differences between participants through the measurement at different times.

5 Conclusion

The combination of BFRT and EMS for low-intensity squat training can improve the muscle strength of lower limbs by

promoting muscle hypertrophy and increasing muscle activation, which makes up for the limitations and deficiencies of each intervention method used alone.

Data availability statement

The raw data supporting the conclusion of this article will be made available by the authors, without undue reservation.

Ethics statement

The studies involving human participants were reviewed and approved by the ethics subcommittee of Chengdu Sport University (CDSUEC_2020_17_57). The patients/participants provided their written informed consent to participate in this study.

Author contributions

NL designed the study, was responsible for the primary data acquisition, assisted with the analysis of the data, and drafted the manuscript. JY assisted with the collection of the data and drafted the manuscript. YL conceived the study and provided significant insights for the direction of this study, and drafted the manuscript. All authors read and approved the final manuscript.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Effects of "accurate measurement" comprehensive sports activities on balance ability, body composition and bone density of female college students

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Background: A sedentary lifestyle with little movement has affected modern youth, and regular exercise has real benefits for people; such studies are mostly for older adults, and more evidence is needed for adolescents.

Objective: To compare differences in balance, body composition, and bone mineral density among female college students before and after an exercise intervention to provide precise evidence that exercise promotes college student health.

Methods: A whole group of female students in a university was sampled and included in the statistical analysis 50 people, divided into two cohorts, 21 people in the test group and 29 people in the control group; the test group had 4 comprehensive sports activities per week and the control group had 1 comprehensive sports activities per week, and the differences in each index of balance ability, body composition and bone density before and after the intervention were compared after 3 months.

Results: After exercise intervention, when maintaining balance, the area of the center of gravity movement trajectory increased by 32.36% in the test group compared with the pre-intervention period and increased by 42.80% compared with the control group, and the differences were all statistically significant ($p < 0.01$); body mass index (BMI), body fat rate (BFR), visceral fat area (VFA), skeletal muscle content, and Inbody score increased over time more reasonable, and the difference in the effect of time factor (effect) was statistically significant ($p < 0.01$); bone mineral density (BMD) and BMD Z value increased with time, and the difference in the effect of time factor was statistically significant ($p < 0.05$).

Conclusion: Female college students' body balance ability improved substantially after exercise intervention; at the university level, female college students had a more rational body composition and continued natural increase in BMD, which were not related to exercise intervention.

KEYWORDS

balance ability, body composition, bone mineral density, exercise intervention, college girl

1 Introduction

With the advancement of technology, people's lifestyles have changed a lot, such as smartphone use, which increases sedentary time and reduces physical activity time, especially for young people (González-ruiz et al., 2018; Guthold et al., 2018; Kemp et al., 2022). Physical inactivity is the cause of many chronic diseases and musculoskeletal disorders, such as sarcopenia and osteoporosis, which place a significant health and social service burden on countries. The international community has recognized physical activity as a priority for health promotion and physical activity as an effective non-pharmaceutical intervention. Human balance is the sum of complex reflexes that coordinate the movement of all parts of the body based on proprioceptive, visual and vestibular sensory information through the regulation of the central nervous system (Jin and Yu, 2005). Balance plays an important role in daily life, not only as one of the main abilities for physical activity, but also as one of the main scientific bases for fall prevention. Body composition is an important indicator to accurately evaluate the health level and body size standard of adolescents, which can monitor their nutritional status and growth and development, and has an important impact on athletic ability; body composition in adolescence has strong plasticity and is a critical period for benign development. Bone mineral density (BMD), is an important indicator of bone health and is used to assess bone strength. The survey shows that the low bone mass rate of people over 50 years old in China is 46.4%, including 46.9% for men and 45.9% for women (Chinese Medical Association Division of Osteoporosis and Bone Mineral Salt Diseases, 2019).

Physical activity history in relation to body composition and functional capacity (Bland et al., 2021; Buckinx et al., 2021). Meta-analytical studies of resistance exercise are effective in improving lower limb muscle strength, increasing lean body mass and reducing body fat in cancer patients (Padilha et al., 2017). One study used 12 weeks of high-intensity interval training to improve physical function and body composition in sedentary obese older adults, with positive results (Buckinx et al., 2019). An analysis of the association between environmental factors and their bone mass in 200 young people showed that body mass index (BMI) and physical activity level were significantly associated with bone health in young people (Correa-rodriguez et al., 2018). Young female athletes participating in high weight-bearing activity sports exhibit higher levels of bone mineral density in the lower extremities (Helder et al., 2018). One study measured higher whole-body bone density in female runners than in female non-runners (Mccormack et al., 2019). 16-week comprehensive exercise training intervention for HIV-infected patients results in increased muscle strength and improved depression and quality of life indices compared to recreational group (Oliveira et al., 2020). Muscle atrophy is characterized by a reduction in muscle mass and strength, which can lead to a number of age-related diseases. The loss of muscle mass can directly affect the quality of daily life and is a major risk factor for chronic diseases (Brunetta et al., 2020). Physical activity is important for maintaining health by promoting muscle protein synthesis and activating signaling pathways that regulate muscle fiber metabolism and function (Shen et al., 2018). Positive correlation between BMD and skeletal muscle (ASM) mass of the extremities in 1,343 men aged 40 years or older in a Chinese community at three-year follow-up (Han et al., 2021). There is a significant positive correlation between muscle mass, bone density and adiposity in elderly women. Appropriate exercise and nutritional support can increase muscle mass, improve bone density and promote the positive effect of adipose tissue, which is conducive to improving balance, preventing the risk of falls, fractures and disability and improving the quality of life of the elderly (Yin et al., 2022). There is an investigation into the characteristics of changes in age, BMI and body composition in postmenopausal women and the relationship between them and osteoporosis. Age is a risk factor for osteoporosis in postmenopausal women, while BMI, muscle mass and fat mass are protective factors that contribute to overall BMD and overall bone mineral content (Yuan et al., 2022a). Analysis of clinical data from 1,032 cases of postmenopausal women in China showed that skeletal muscle index of the extremities was negatively associated with the risk of osteoporotic fracture at major sites and the risk of hip fracture (Lin et al., 2022a). Postmenopausal women's bone mineral density is closely related to changes in skeletal muscle mass (Zhao et al., 2011). Women are more likely to suffer from osteoporosis, and body composition can be used as an indirect reference indicator to evaluate bone health status (Yuan et al., 2022b). Study finds positive health effects of team sports training for sedentary women (Møller et al., 2020).

In review, studies on the effects of exercise on physiological function, muscle and bone density have focused on areas related to promoting recovery from disease (Varahra et al., 2018; Schootemeijer et al., 2020; Chen et al., 2022; Mavropalias et al., 2022; Talotta et al., 2022), or for middle-aged and elderly populations (Du et al., 2022; Papp et al., 2022), or for postmenopausal women (Shojaa et al., 2020; Hejazi et al., 2022), studies on adolescents are relatively fewer. However, the health problems caused by sedentary behavior of college students cannot be ignored. Female college students are in late adolescent development, their body physique is gradually becoming adult and fixed, and their health status is of great significance for future work and childbirth. Before and after puberty is a critical stage of bone density development, and obtaining the highest possible bone mass plays an important role in preventing the occurrence of osteoporosis (Zhang et al., 2000). Late adolescence is the 2nd critical period for maintaining bone mass, lasting 2–5 years, during which appropriate physical training can help maintain peak bone mass (Li et al., 2012). Physical exercise strengthens the muscles and bones as a whole unit and can improve muscle strength and body balance. There is a correlation between muscle activity and bone density (Calendo et al., 2014; Cidem et al., 2014). Muscles and bones are closely related, the two complement each other, scientific exercise will play a dual effect (Zhao et al., 2017). At this stage, the myopia rate of Chinese school students is rapidly rising and overweight is increasing year by year. Some scholars suggest that "accurate measurement", "accurate analysis" and "accurate intervention" of students' physical fitness is the key to solve the development of students' physical health (This publication reports, 2020). Therefore, to conduct exercise intervention for female college students, professional intervention personnel, precise instruments, and specialized testers were used to accurately measure three aspects of balance ability, body composition, and BMD, and to compare the differences of each index before and after the intervention. School physical education

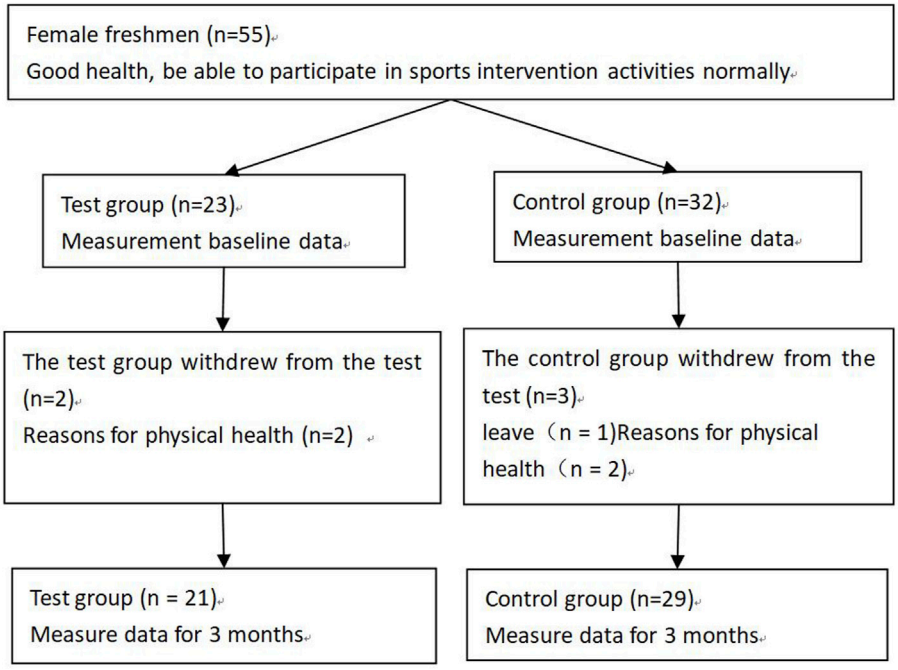

FIGURE 1
Flowchart of intervention process.

classes generally include a variety of physical exercises to improve overall physical fitness, and this study selected integrated physical activities that included running, jumping, and throwing for the intervention. It is hypothesized that comprehensive sports activities has a good effect on balance ability, body composition, and BMD, providing more evidence for female college students' exercise for health.

2 Materials and methods

2.1 Subjects and exercise intervention

The subjects were female college students, and the sample size was estimated by G*Power3.1.7. Inclusion Criteria: Healthy and normal participation in sports intervention activities. Exclusion criteria: unable to complete physical activities as planned and unable to participate in the test on time. A whole cohort sample of female students in two first-year university classes at a general university, with 23 in the test group (class) and 32 in the control group (class), for a total of 55 students. The intervention was organized 4 times a week for 100 min of comprehensive sports activities in the test group and 1 time a week for 100 min of comprehensive sports activities in the control group, with a 3-month follow-up. Comprehensive sports activities include running, jumping, throwing, coordination and flexibility exercises with jogging, stretching, aerobics, basketball activities, standing long jump activities, *etc.* The intervention was implemented by university physical education teachers at school teaching sites with 90% or more attendance. The intervention flow chart is shown in Figure 1. All subjects signed a written informed consent form. The study was approved by the Academic Ethics Committee of Heze University. The study was registered with the Chinese Clinical Trials Registry under the number ChiCTR2000039041.

2.2 Tools and data acquisition

The balance ability test was performed using the Balance Function Training and Assessment System (Model: XY-PH-111, China). The subjects stood barefoot on the test platform, calibrated the footprint position according to the display in front of them, made the center of gravity cast point at the cross center marked on the screen, and tilted their bodies to the limit of stability in 8 directions: forward, right front, right, right back, back, left back, left and left front according to the screen arrow prompts. The subject observes his or her center of gravity on the display and tilts as far as possible in the direction of maximum tilt for more than 2 s. During the whole test, if the subject's feet move or fall, the test is repeated. The tester is mainly composed of three parts: pressure sensor (test platform), computer and application software. The pressure sensor can record the swaying of the body's center of gravity, and convert the recorded signal into data input to the computer, the computer, with the support of application software, analyzes the received data, describes the projection trajectory of the body's center of gravity on the pressure plate in real time, and automatically records the trajectory of the body's center of gravity movement, the area formed by the trajectory (balance area) can record a very small amount of center of gravity movement, which can be compared quantitatively The

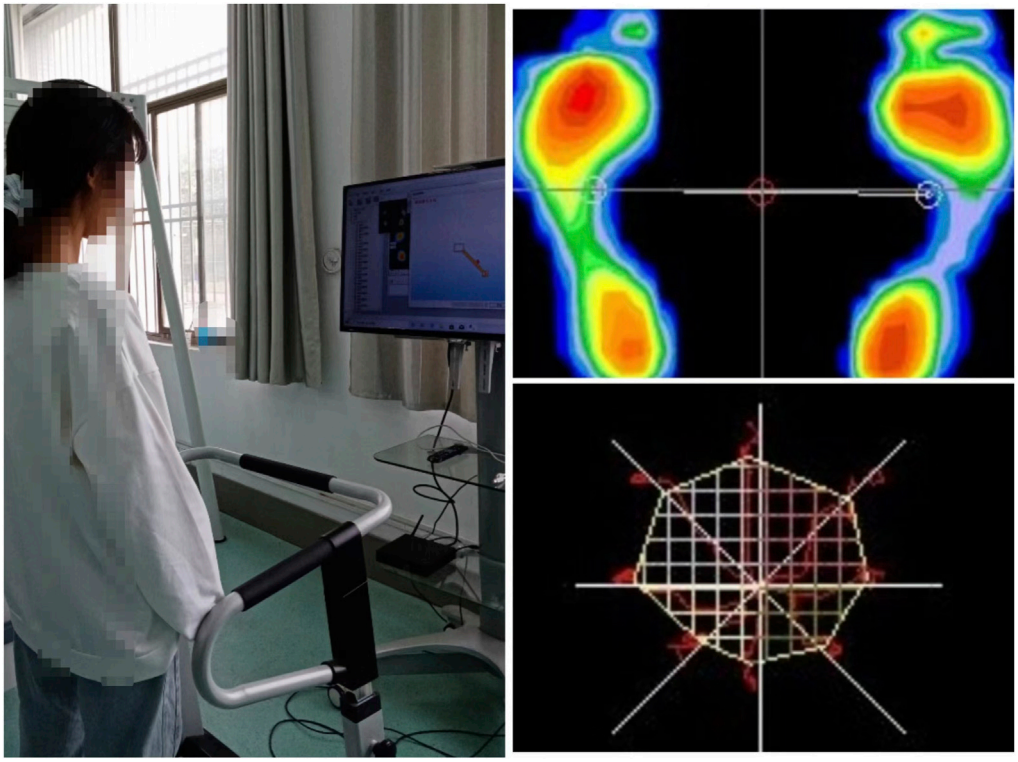

FIGURE 2
Balance test.

area formed by the trajectory (balance area) can be recorded to a very small amount of center of gravity movement, which can reflect the balance function in a more quantitative and objective way and facilitate comparison between different testers. See Figure 2.

Body composition testing was performed using the Biospace body composition analyzer (model: InBody 770, South Korea). Using bioelectrical impedance analysis (BIA), a non-invasive method that involves placing electrodes on the feet and hands, low-frequency electric current is passed through the entire body, and the flow of current will be affected by body water. The BIA device measures how the current signal is obstructed by different types of tissue (muscle has high conductivity, but fat can slow down the signal), and after the BIA determines the resistance to the flow of current through the body, body fat and other body composition data can be calculated using an equation based on the value of body water. Body composition was selected from five indicators that are closely related to the physical health of female university students. Body Mass Index (BMI), BMI (kg/m^2) is a standard commonly used internationally to measure the degree of body fat and thinness and whether it is healthy or not, and the calculation formula is: BMI = weight ÷ $height^2$; Body Fat Rate (BFR) is the proportion of body fat weight in the total body weight (%), which reflects the amount of fat content in the human body. Visceral fat area (VFA) is the area of visceral fat (cm^2) in CT image of abdominal belly button section, which is one of the test indicators of body composition analyzer and is an important indicator to evaluate whether it is hidden obesity. Skeletal muscle content (kg) directly reflects the body's activity function, and any activity is the result of skeletal muscle contraction. The Inbody score is based on the difference between muscle mass and fat mass measurements and the standard value. The base score is 80 points, and the muscle mass measurement is lower than the standard value.

Bone density testing was performed using an OsteoSys dual-energy X-ray bone densitometer (model: EXA-3000, South Korea). The left ulnar flexure (forearm) distal bone mineral density (BMD) and Z value were tested; Z value indicates the ratio of the subject's BMD to the average peak BMD of the same age, the same sex, and the same race, and the Z value can be used to understand the position of the subject's BMD compared to that of the same age, indicating a higher or lower BMD value than that of people of the same sex, age, and race. The testers are specially trained and the tests are conducted in specialized laboratories.

2.3 Statistical analysis

The data were imported into SPSS20.0 statistical software. Descriptive statistics were used for the basic situation of the subjects, and independent sample *t*-test was used for comparison between different groups of baselines. When examining the effect of exercise on the balance ability, body composition and bone mineral density of female college students, the repeated measurement analysis of variance of 2 (test time: pre-test, post-test) × 2 (group: test group, control group) was used to compare the interaction within the group (pre-test, post-test) and between groups. The spherical test was carried out before the repeated

TABLE 1 Comparison of basic conditions between the test group and the control group (n = 50, $\bar{x} \pm S$).

Subjects	Test group (n = 21)	Control group (n = 29)	t	P
age (year)	19.48 ± 0.81	19.24 ± 0.64	1.15	0.26
height (cm)	163.47 ± 4.80	162.79 ± 7.20	0.37	0.71
weight (kg)	58.11 ± 8.70	59.74 ± 12.17	−0.52	0.60

TABLE 2 Comparison of balance area between the two groups of college students before and after the experiment (n = 50, $\bar{x} \pm S$).

	Test group (n = 21)		Control group (n = 29)		Time F value (η^2)	Time × group F value (η^2)
	Pretest	Posttest	Pretest	Posttest		
balance area (mm^2)	8,696 ± 2,435	11,510 ± 3,666	8,188 ± 3,153	8,060 ± 2,729	8.06 (0.14)*	9.68 (0.17)**

Note: *$p < 0.05$, **$p < 0.01$.

measurement analysis of variance, and the Greenhouse-Geisser method was used when the football shape was not satisfied. If the interaction was significant, the Bonferroni correction was used for post-test, and the intra-group and inter-group differences were compared through simple effect analysis. In the statistical analysis, in order to reflect the test effect, the effect quantity index η^2 is reported, which is considered to be a small effect when $\eta^2 = 0.01$, a medium effect when $\eta^2 = 0.06$, and a large effect when $\eta^2 = 0.14$. When $p < 0.05$, the difference was statistically significant.

3 Results

3.1 Basic information of the subjects

A total of 50 female college students were included in the statistical analysis, including 21 in the test group and 29 in the control group (Table 1). It is considered that the baseline of the test group and the control group is balanced.

3.2 The influence of exercise on the balance ability of female college students

The differences were not statistically significant in the independent sample t-test between the test and control groups for the premeasured balance area. (t = 0.62, p = 0.54). A 2 × 2 repeated-measures ANOVA was used to determine the difference in the change of human balance area over time between the test and control groups (Table 2). Because the interaction between group and time had a statistically significant effect on the balance area ($p < 0.0L$), the simple effect analysis of the two factor groups and time showed that the post-test in the test group increased 2814 mm^2 by 32.36%, and the difference was statistically significant ($p < 0.0L$). The post-test in the control group decreased 128 mm^2 compared with the pre-test, but the difference was not statistically significant (p = 0.84); In the post-test, the test group had more 3450 mm^2 than the control group, an increase of 42.80%, and the difference was statistically significant ($p < 0.01$).

3.3 The influence of exercise on the body composition of female college students

Body composition is closely related to body mass index (BMI), body fat rate (BFR), visceral fat area (VFA), skeletal muscle content and Inbody score. There was no significant difference in BMI, BFR, VFA, skeletal muscle content and Inbody score between the test group and the control group ($p > 0.05$). A 2 × 2 repeated-measures ANOVA was used to determine the differences in the changes of each index between the test and control groups over time (Table 3).

3.4 Effect of exercise on bone mineral density of female college students

There was no significant difference in bone mineral density (BMD) and BMD Z value between the test group and the control group ($p > 0.05$). A 2 × 2 repeated-measures ANOVA was used to determine the differences in the change of each index between the test and control groups over time (Table 4).

4 Discussion

The results of the study female college students after exercise intervention, the balance area of the test group increased substantially, and the control group decreased slightly, proving that exercise is important for enhancing the balance ability of female college students. Studies have demonstrated that exercise is beneficial in improving functional capacity and dynamic balance in children with hemiplegia (Soliman et al., 2022), a randomized controlled trial showed that a gait training program used by stroke patients was beneficial in improving balance (Shaw et al., 2022), and another study female university students through 8 weeks of fitness qigong exercises. The result was a significant increase in balance test

TABLE 3 Comparison of body composition indicators between the two groups of college students before and after the experiment (n = 50, $\bar{x} \pm S$).

	Test group (n = 21)		Control group (n = 29)		Time F value (η^2)	Time×group F value (η^2)
	Pretest	Posttest	Pretest	Posttest		
BMI(kg/m^2)	21.74 ± 2.75	21.39 ± 2.52	22.35 ± 3.23	22.06 ± 3.48	9.80 (0.17)**	0.07 (<0.01)
BFR(%)	31.94 ± 5.92	29.73 ± 5.67	33.81 ± 4.14	30.99 ± 4.83	87.77 (0.65)**	1.31 (0.03)
VFA (cm^2)	87.35 ± 30.53	76.78 ± 26.52	99.01 ± 34.83	86.04 ± 35.79	59.19 (0.55)**	0.61 (0.01)
Skeletal muscle (kg)	21.03 ± 2.70	21.39 ± 2.43	21.03 ± 3.60	21.61 ± 3.68	24.02 (0.33)**	1.49 (0.03)
Inbody score	69.29 ± 5.07	71.48 ± 4.84	68.45 ± 3.28	71.34 ± 3.60	80.51 (0.63)**	1.55 (0.03)

Note: **$p < 0.01$.

TABLE 4 Comparison of bone mineral density between the two groups of college students before and after the experiment (n = 50, $\bar{x} \pm S$).

	Test group (n = 21)		Control group (n = 29)		Time F value (η^2)	Time × group F value (η^2)
	Pretest	Posttest		Pretest		
BMD(g/cm^3)	0.43 ± 0.07	0.44 ± 0.07	0.41 ± 0.05	0.44 ± 0.06	6 (0.11)*	0.20 (<0.01)
Z (g/cm^3)	−0.79 ± 1.01	−0.51 ± 1.24	−1.05 ± 0.72	−0.64 ± 0.94	5.64 (0.11)*	0.19 (<0.01)

Note: *$p < 0.05$.

scores post-intervention compared to pre-intervention scores (Su et al., 2018). These are similar to the results of this study, and the fact that healthy female college students can consistently improve their balance through comprehensive sports activitiesis an exciting and noteworthy result. Balance is an important physiological function of the human body, which is the ability to maintain the posture of the body; especially the ability to control the body's center of gravity on a small support surface. Balance is an important safeguard for the body to maintain standing, walking and coordinating various movements. The postural balance of the human body depends on the coordination of visual, proprioceptive and vestibular information and the control of motor effectors by the central system. Information from the visual system is collected by the retina and transmitted via the visual pathway to the visual center, providing information about the surrounding environment and the body's movement and direction; proprioception transmits information about the state of muscles, joints, tendons and other effector organs; and vestibular sensation is the main structure for maintaining balance and perceiving the body in relation to the surrounding environment. The main manifestations of the human body when balance function is impaired include: low muscle strength and endurance; decreased flexibility of joints and soft tissue flexibility; dysfunction of the central nervous system; decreased visual, vestibular function, and proprioceptive efficiency; reduced tactile input and sensitivity; and reduced spatial perception (You and Wen, 2014)In the field of clinical research, balance testing and training tools have been recognized and accepted by many people as an adjunct to the detection and treatment of patients with balance disorders.

In clinical practice, the most commonly used body composition analysis is measured using the bioelectrical impedance analysis (BIA) method (Zhao and Li, 2016). This method takes advantage of the fact that different components of the human body have different water content and different electrical conductivity, and combines data such as height, weight, gender, and age to estimate the proportion of different components of the human body such as muscle and fat. The differences in the effects of the time factor on BMI, body fat percentage, visceral fat area, skeletal muscle, and Inbody score in this trial were all statistically significant, with reductions in BMI, BFR, and VFA, increases in skeletal muscle, and Inbody score of varying magnitude, and an overall more reasonable body composition, but the results showed that these differences were not related to the exercise intervention factors. The musculoskeletal benefits of exercise are clearly beneficial for middle-aged and older adults (Lim and Goh, 2022; Schaun et al., 2022), and studies have concluded that comprehensive content exercise helps by improving muscle size, muscle strength, balance and physical function in older adults that have decreased because of age (Shahtahmassebi et al., 2022). Due to changing lifestyles and increased academic stress, physical and mental health problems among college students have become increasingly serious, and there are systematic evaluations of qigong exercises considered to improve physical and mental health of college students, with no significant effects on muscle strength found (Lin et al., 2022b). There is also a study of the effects of 8 weeks of resistance training on physical performance and body composition in adolescent athletes, with significant benefits in muscular endurance, strength and static balance (Fayazmilani et al., 2022). These exercises had no statistically significant changes in muscle and effects on balance similar to the present study. It is possible that the age stage or the lifestyle of college students led to a small effect of the intervention in body composition.

Dual-energy X-ray bone densitometry is the most widely used bone densitometry technique at present. In this study, the BMD and BMD Z-value post-test and control group both increased compared with the pre-test, which was statistically significant and may be

related to the fact that the BMD of college students is still in a natural growth state at this stage. The interaction of BMD content, BMD Z-value group and time between the test and control groups was not statistically significant, and the differences were not related to the exercise intervention factors. In a 2-year study of healthy young women, exercisers did not have an effect on bone mineral density in the ulnar radius, lumbar spine, and proximal femur after 2 years of physical activity (Mazess and Barden, 1991), which is similar to the results of the present study. Although it is generally accepted that physical activity contributes to bone health and thus reduces fracture risk, a review of the medical literature suggests that, paradoxically, road cycling has a negative impact on bone strength and is therefore a risk factor for proximal femur fractures (Simonovich et al., 2022). However, physical activity is known to have anabolic effects on bone tissue, and it has been shown to increase BMD in young people and adolescents (Courteix et al., 1998). Some studies have shown that the development of body systems is basically mature at the end of puberty and bone mass is in a slow growth period, but the bone density increases under the stimulation of resistance training, indicating that suitable load stimulation at this stage can still promote bone density (Li et al., 2019). Because high-intensity resistance exercise produces a powerful stimulus to bone through muscle pulling contraction, the bone produces adaptive changes after this stimulus, promoting enhanced activity of osteoblasts, making bone formation greater than bone resorption, bone mineral accumulation, and eventually bone mass and bone density both increase. The effect of the exercise intervention in this trial was not statistically significant, which may be related to the design of the exercise intervention content, or to the short duration of the exercise intervention, or because the effect of the exercise intervention was not significant at this stage. The cumulative effect of physical activity on BMD must have a certain amount of time to manifest itself, and many factors must be taken into account in the effect of physical activity on BMD, such as type of exercise, intensity of exercise, frequency of exercise, duration of each exercise, and duration of exercise intervention (Qu, 2020). Studies have shown a correlation between grip strength and free fat mass and bone mineral density (Pelegrini et al., 2022), and physical activity capacity was positively associated with hip BMD in both young women and older men, and although it is unlikely to increase BMD, the findings suggest that maintaining a high level of physical fitness and activity may help prevent bone loss in middle-aged and older adults (Hauger et al., 2020), and 8 weeks of power cycling high-intensity interval training easily improved physical form in overweight female college students and increased femoral bone mineral density in normal weight female college students (Li et al., 2022). Therefore, adolescents need to reduce body fat and safeguard dietary nutrition through scientific and regular exercise to improve overall bone density (Xu et al., 2021). Muscle and bone are anatomically and physiologically connected and together have a regulatory effect on body function. There are few studies on the association between physical function, body composition and bone mineral density in adolescents, and more prospective studies are needed in the future to provide more reference indicators for peak physical function and peak bone mass. The limitations of this study are mainly in the lack of representativeness of the whole group sampling of girls.

5 Conclusion

After a comprehensive exercise intervention, the balance ability of female university students was improved significantly. Female college students showed decreases in body composition BMI, body fat percentage, and visceral fat area over time at the university level, with varying increases in skeletal muscle content and Inbody score, and a more reasonable body composition overall; female college students showed increases in bone mineral density content and bone mineral density Z-score over time; these were not related to the comprehensive exercise intervention. The effect of the combined exercise intervention on body composition and bone mineral density in female college students was not statistically significant, which may be related to age and environmental factors in female college students, and may also be related to the type of intervention.

Data availability statement

The original contributions presented in the study are included in the article/Supplementary Material, further inquiries can be directed to the corresponding author.

Ethics statement

The studies involving human participants were reviewed and approved by Academic Ethics Committee of Heze University, Heze University. The patients/participants provided their written informed consent to participate in this study.

Author contributions

ZZ designed the paper and wrote the manuscript, JEL and JGL participated in the data collation, and JPL participated in the experimental design. All the authors contributed to this article and approved the submitted version.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Test–retest reliability of kinematic and kinetic parameters during dual-task stair walking in the elderly

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Objective: This study aims to evaluate the test–retest reliability of kinematics and kinetics during single and dual-task stair walking in the elderly.

Methods: Fifteen healthy elderly adults were recruited. Kinematic and kinetic parameters were measured using an infrared motion analysis system (Vicon, Oxford Metrics Ltd., Oxford, United Kingdom) and force platforms (Switzerland, Kistler 9287BA and 9281CA). Participants were tested under single-task and dual-task (serial 3 subtractions or carrying a cup of water) conditions. Each participant completed two sessions on two separate days with a 1-week interval. Intraclass correlation coefficients (ICC), Pearson correlation coefficient (r), and Bland–Altman plot were used to assess the reliability of stair walking.

Results: When ascending stairs, the ICC of kinematics and kinetics ranged from fair to excellent (ICC = 0.500–0.979) in the single and dual tasks, except for step length (ICC = 0.394) in the single task. The r value of kinematics and kinetics ranged from 0.704 to 0.999. When descending stairs, the ICC of kinematics and kinetics ranged from good to excellent (ICC = 0.661–0.963), except for min hip moment (ICC = 0.133) and min ankle moment (ICC = 0.057) in the manual task. The r value of kinematics and kinetics ranged from 0.773 to 0.960 in the single and dual tasks. In the Bland–Altman plots, all the zero values and most of the dots fell in the 95% confidence interval, and the mean difference was found to be close to zero for all the parameters during stair walking.

Conclusion: These results obtained from this study show the good test-retest reliability of step cadence, step speed, and step width during single- and dual-task stair walking in the elderly, and the poor reliability of step length during ascending stairs. All the kinetic parameters, including min hip moment, max knee moment, and min ankle moment, had good test-retest reliability during single- and dual-task stair walking, but min hip moment and min ankle moment had poor reliability during manual-task descending stair. These results may help researchers in the assessment of biomechanics of dual-task stair walking in the elderly and to interpret the effect of interventions in this population.

KEYWORDS

dual-task, stair walking, reliability, kinematics, kinetics

1 Introduction

Approximately 30% of elderly people aged 65 years or older fall each year (Chippendale and Raveis, 2017). Falling during stair walking is a common cause of musculoskeletal injuries, hip fractures (Kim et al., 2022), craniocerebral injury, and even mortality (Clark and Arnold, 2021; Kim and Kim, 2022), leading to a substantial economic burden for family and society (Moreland et al., 2020). Stair walking is a difficult and challenging task for elderly people due to the deterioration of their physical function (Ito et al., 2020). More than half of all reported fall injuries occur when walking on stairs or slopes (Sheehan and Gottschall, 2012) (Blazewick et al., 2018).

The test–retest reliability of kinematic and kinetic parameters during single-task stair walking or level walking has been demonstrated in literature (Mentiplay and Clark, 2018); (Jarvis et al., 2021); (Leitner et al., 2011). As a previous study demonstrated, the test-retest reliability of all gait models was mostly good to excellent during level walking (Mentiplay and Clark, 2018). And the test-retest reliability of walking speed is excellent (ICC = 0.93–0.94) when using the Gait Box during level walking (Jarvis et al., 2021). Under a single task, stair walking had moderate to good reliability, with intraclass correlation coefficients (ICC) ranging from 0.537 to 0.836 and coefficients of variation (CV) ranging from 2.51% to 6.51% in vertical ground reaction forces (Leitner et al., 2011). In the single-task walking condition, all the components of the measure suggest good to excellent reliability (Brusola et al., 2021).

Dual-task interference could further impair gait performance and increase the risk of fall during stair walking in older people (Vallabhajosula et al., 2015). The dual-task (DT) paradigm leads to the competition of attention resources available and will influence the posture control of the elderly in stair negotiation (Simoni et al., 2013; Bianca et al., 2021). This paradigm refers to the performance of a cognitive task (CT) and manual task (MT) simultaneously, such as talking, calculating, and texting while moving an object to another place or carrying a tray with cups of water during stair walking (Paul et al., 2009; Asai et al., 2014; Belur et al., 2020). Dual tasks could significantly decrease gait performance and lower limb joint movement (Wang et al., 2022), reduce the gait speed, and increase the step width and body sway during stair walking (Madehkhaksar and Egges, 2016; Zhang et al., 2018).

However, the reliability of stair walking under a dual-task condition in the elderly is currently unknown. This study aimed to assess the test–retest reliability of kinematic and kinetic parameters in stair negotiation under single and dual-task conditions. Results may serve as a basis for clinical assessment of gait and motor control during stair walking in older people. We hypothesize that the 1) the reliability of kinematic and kinetic parameters of stair walking is high under single-task conditions 2) and poor under dual-task conditions.

2 Materials and methods

2.1 Participants

Fifteen healthy elderly adults (7 women and 8 men, 66.8 ± 3.6 years old, 161 ± 4.3 cm, 61.6 ± 4.2 kg) were recruited from a local community by distributing flyers in Jinan city, China. Some studies have reported that the estimation of the minimum sample size can be calculated based on the reliability of the ICC and the number of measurements of the subject (Donner and Eliasziw, 1987). A previous study found that the test-retest reliability was good to excellent for most models for lower limb kinematics and kinetics (ICC≥0.8) (Mentiplay and Clark, 2018). We set ICC = 0.8, α = 0.05, and power = 0.8, and when the number of measurements of the subject is two, according to the model provided by Donner A, the minimum sample size should be fifteen (Donner and Eliasziw, 1987). All the participants signed an informed consent form. This study was approved by the medical ethics committee of Shandong Sport University (No. 2017103). Participants met the following criteria for inclusion: able to ascend and descend stairs independently; able to understand verbal and written information; medically stable with no other diseases that significantly influenced gait performance (Flansbjer et al., 2005); and Body Mass Index of 18.5–25 kg/m^2. Exclusion criteria included lower extremity contractures or pain (Dobkin et al., 2011); musculoskeletal or cardiovascular pathologies (Hashish et al., 2017); and Mini Mental State Examination score (MMSE) of <24 points (Lopez et al., 2005).

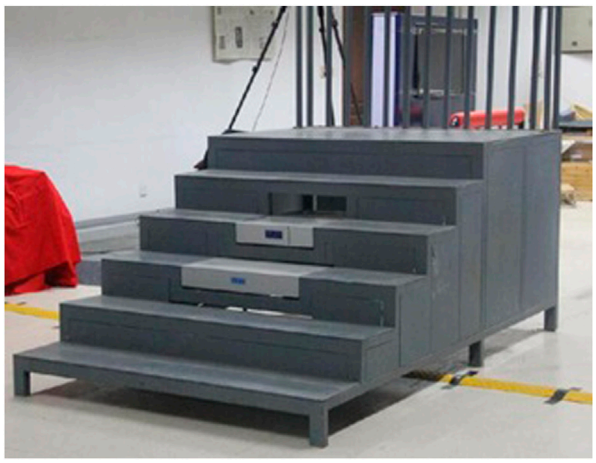

FIGURE 1
The figure of simulated stairs.

2.2 Data collection

Data were collected in the Biomechanical Laboratory of the Shandong Sport University. Each participant completed the stair walking test in two sessions on two separate days with a one-week interval. The tests were performed by the same experimenter at the same time of day in the same laboratory. A simulated staircase (17 cm riser, 29 cm tread) with six steps was constructed for data collection, which met the national residential standards (Figure 1). Two force platforms (Switzerland, Kistler 9287BA and 9281CA, 90 cm × 60 cm×10 cm) were used to collect kinetic data at 1,000 Hz and embedded in the third and fourth steps of the simulated staircase (Komnik et al., 2018). An infrared motion analysis

system with 12 cameras (Vicon, Oxford Metrics Ltd., Oxford, United Kingdom) was used to collect kinematic data at 100 Hz (Komnik et al., 2018; Bates et al., 2021). Kinematic and kinetic data acquisition was synchronized using Vicon Nexus software. Prior to data collection, the participants were asked to wear unified black tight-fitting clothes and shoes and anthropometric data were collected (Law et al., 2021). The dominant leg was determined as the preferred leg for kicking a football. Forty-one reflective markers (14 mm) were placed on each participant's anatomical landmarks based on the Visual 3D model, including four markers on the head, two on the trunk, 12 on the upper extremities, five on the pelvis, and 18 on the lower extremities. All the participants were instructed to ascend and descend the staircase with the dominant foot for the first step in a step-by-step manner at a comfortable pace under single-task, cognitive task, and manual-task conditions in random order. In the single-task condition, the participants descend and ascend stairs without cognitive or manual task interference. For the cognitive-task condition, the participants descend and ascend the stairs while performing subtraction of serial threes from a random three-digit number. For the manual-task condition, the participants descend and ascend the stairs while carrying a glass of water (0.63 kg) with the dominant hand. No instruction to prioritize the tasks were given to eliminate interference. Five trials were collected for each condition with a 1-min break between consecutive trials.

2.3 Data processing

When ascending the stairs, a stride cycle was defined as the first foot contact on the second step and ended at the same foot contact on the fourth step. When descending the stairs, the selected stride cycle started with the foot contact (of the same foot) on the fourth step and ended with the foot contact (of the same foot) on the second step (Protopapadaki et al., 2007). The first double-limb support phase was defined as when the right foot touched down until the left foot took off (Gorelick et al., 2009; Buckley et al., 2018). The single-limb support phase was defined as when the left foot took off until the left foot touched down. The second-double limb support phase was defined as when the left foot touched down until the right foot took off. The swing phase was defined as when the right foot took off until the right foot touched down (Singhal et al., 2014; Quintero et al., 2015). Vicon Nexus data were used to command, denoise the infrared reflective markers, and delete the markers in static and dynamic acquisition data. We completed the data by interpolation and intercepted the action frame number according to the gait cycle definition, renamed the intercepted data, and exported it to the C3d file. The exported static and dynamic c3d files were imported to Visual 3D for in-depth data processing and index extraction. A static model was built using the files and connected with the dynamic data. After selecting the data, the kinematic and kinetic signals were low-pass filtered by Pipeline with a cutoff frequency of 6 Hz (Singhal et al., 2014).

2.4 Parameters

2.4.1 Kinematic parameters

Step cadence was expressed as the number of steps in a minute (steps/min). Step speed was obtained by dividing the stride cycle duration into the displacement of the center of mass (m/s) (Buckley et al., 2018). Step length was expressed as the distance between the heel of the right foot during stair walking from the landing of the right foot to the landing again(m). Step width was defined as the medial–lateral distance between the midpoints of each foot(m) (Baudendistel et al., 2020).

2.4.2 Kinetic parameters

The net moment of the hip–knee–ankle joint was obtained based on the inverse dynamics calculation formula. Specific settings were used in V3d software to select the mechanical parameters to be calculated, the lower limb joint, and the reference link to complete the calculation of the net moment of the lower limb joint. In this study, the flexion and extension moments of the joint in the sagittal plane around the *x*-axis were selected for analysis. Joint movement was expressed as the peak flexion and extension moment of the hip, knee, and ankle joint in the sagittal plane (N·m/kg).

2.5 Statistical analysis

The kinematic and kinetic data were analyzed by SPSS (version 25.0) and presented as mean and standard deviation. Test—retest relative reliability was assessed using the intraclass correlation coefficient (ICC) test and interpreted as follows: poor ($0 <$ ICC <0.39), fair ($0.40 \leq$ ICC <0.59), good ($0.60 \leq$ ICC < 0.74), and excellent ($0.75 \leq$ ICC < 1.0) (Cicchetti and Sparrow, 1981). Pearson correlation analysis was used to assess the linear correlation of the data from the two tests. The relationship was assessed as: small (0–0.3), moderate (0.3–0.5), significant (0.5–0.8), and highly relevant (0.8–1.0). The Bland–Altman method was used to assess the distribution of the mean and the difference of the two test data sets by using a scatter plot of the differences (Aandstad and Simon, 2013). Statistical significance was set at $p < 0.05$ (Clemons and Harrison, 2008).

3 Results

3.1 Test-retest reliability of kinematic and kinetic parameters during ascending stairs

Table 1 shows that the interclass correlation coefficient values ranged from 0.909 to 0.963 in step cadence, from 0.866 to 0.942 in step speed, from 0.394 to 0.553 in step length, and from 0.511 to 0.773 in step width under the three task conditions. The relationship values ranged from 0.840 to 0.929 in step cadence, from 0.768 to 0.891 in step speed, from 0.203 to 0.246 in step length, and from 0.344 to 0.630 in step width under the three conditions.

Table 2 shows that the interclass correlation coefficient values ranged from 0.858 to 0.922 in min hip moment, from 0.881 to 0.954 in maximum knee moment, and from 0.857 to 0.979 in minimum ankle moment under the three task conditions. The relationship values ranged from 0.751 to 0.856 in minimum hip moment, from 0.787 to 0.914 in maximum knee moment, and from 0.764 to 0.960 in minimum ankle moment under the three conditions.

TABLE 1 The test-retest reliability of kinematic parameters under three conditions when ascending stairs.

		First session	second session	ICC	95% CI	r	P
Single task	Step cadence (step/min)	97.4 ± 11.5	98.0 ± 9.8	0.936	0.870–0.969	0.894	<0.001
	Step speed (m/s)	0.45 ± 0.05	0.47 ± 0.05	0.866	0.726–0.935	0.768	<0.001
	Step length(m)	0.58 ± 0.02	0.59 ± 0.02	0.394	−0.242–0.704	0.246	0.175
	Step width(m)	0.07 ± 0.03	0.08 ± 0.04	0.511	−0.002–0.761	0.344	0.054
Cognitive task	Step cadence	83.9 ± 11.6	83.5 ± 13.2	0.909	0.813–0.955	0.840	<0.001
	Step speed (m/s)	0.40 ± 0.06	0.40 ± 0.07	0.907	0.809–0.954	0.840	<0.001
	Step length(m)	0.58 ± 0.01	0.58 ± 0.01	0.553	−2.181–0.242	0.219	0.229
	Step width(m)	0.09 ± 0.04	0.07 ± 0.04	0.773	0.535–0.889	0.630	<0.001
Manual task	Step cadence (step/min)	85.2 ± 12.4	85.99 ± 12.35	0.963	0.924–0.982	0.929	<0.001
	Step speed (m/s)	0.40 ± 0.05	0.41 ± 0.05	0.942	0.881–0.972	0.891	<0.001
	Step length(m)	0.57 ± 0.01	0.58 ± 0.02	0.500	−2.072–0.268	0.203	0.266
	Step width(m)	0.08 ± 0.05	0.07 ± 0.04	0.741	0.470–0.874	0.594	<0.001

TABLE 2 The test-retest reliability of kinetic parameters under three conditions when ascending stairs. (N.m/kg).

		First session	second session	ICC	95% CI	r	P
Single task	Min hip moment	0.91 ± 0.17	0.90 ± 0.17	0.872	0.738–0.938	0.773	<0.001
	Max knee moment	0.72 ± 0.18	0.72 ± 0.19	0.881	0.755–0.942	0.787	<0.001
	Min ankle moment	1.25 ± 0.29	1.25 ± 0.28	0.979	0.957–0.990	0.960	<0.001
Cognitive task	Min hip moment	0.91 ± 0.17	0.91 ± 0.17	0.858	0.709–0.931	0.751	<0.001
	Max knee moment	0.70 ± 0.18	0.70 ± 0.16	0.954	0.906–0.978	0.914	<0.001
	Min ankle moment	1.21 ± 0.15	1.21 ± 0.15	0.889	0.773–0.946	0.800	<0.001
Manual task	Min hip moment	0.81 ± 0.15	0.78 ± 0.15	0.922	0.840–0.962	0.856	<0.001
	Max knee moment	0.82 ± 0.19	0.83 ± 0.18	0.917	0.830–0.959	0.848	<0.001
	Min ankle moment	1.21 ± 0.18	1.20 ± 0.15	0.857	0.706–0.930	0.764	<0.001

Figure 2 shows the Bland—Altman plots of the kinematic and kinetic parameters when ascending stairs. The systematic basis around the zero line for kinematic and kinetic parameters was revealed. All the zero values fell in the 95% confidence interval. The numbers of dots inside the range of 95% CI interpreted the reliability. A higher number of dots indicated better reliability. In this study, the numbers of dots outside the range of 95% CI were one dot (3.1%) in step width to two dots (6.3%) in step cadence in the single task, three dots (9.4%) in step cadence in the cognitive task, and one dot (3.1%) in step cadence in the manual task. The numbers of dots outside the range of 95% CI were one dot (3.1%) in step speed to two dots (6.3%) in step length in the single task, from one dot (3.1%) in step speed to two dots (6.3%) in step length in the cognitive task, and one dot (3.1%) in step speed and step length in the manual task. The numbers of dots outside the range of 95% CI were two dots (6.3%) in minimum hip moment to three dots (9.4%) in maximum knee moment in the single task, two dots (6.3%) in minimum hip moment in the cognitive task, and one dot (3.1%) in minimum hip moment to two dots (6.3%) in maximum knee moment in the manual task. The numbers of dots outside the range of 95% CI were one dot (3.1%) in minimum ankle moment in the single task, two dots (6.3%) in minimum ankle moment in the cognitive task, and one dot (3.1%) in minimum ankle moment in the manual task.

3.2 Test—retest reliability of kinematic and kinetic parameters during descending stairs

Table 3 shows that the interclass correlation coefficient values ranged from 0.925 to 0.956 in step cadence, from 0.779 to 0.932 in step speed, from 0.305 to 0.745 in step length, and from 0.661 to 0.795 in step width under the three task conditions. The relationship values ranged from 0.866 to 0.916 in step cadence, from 0.639 to

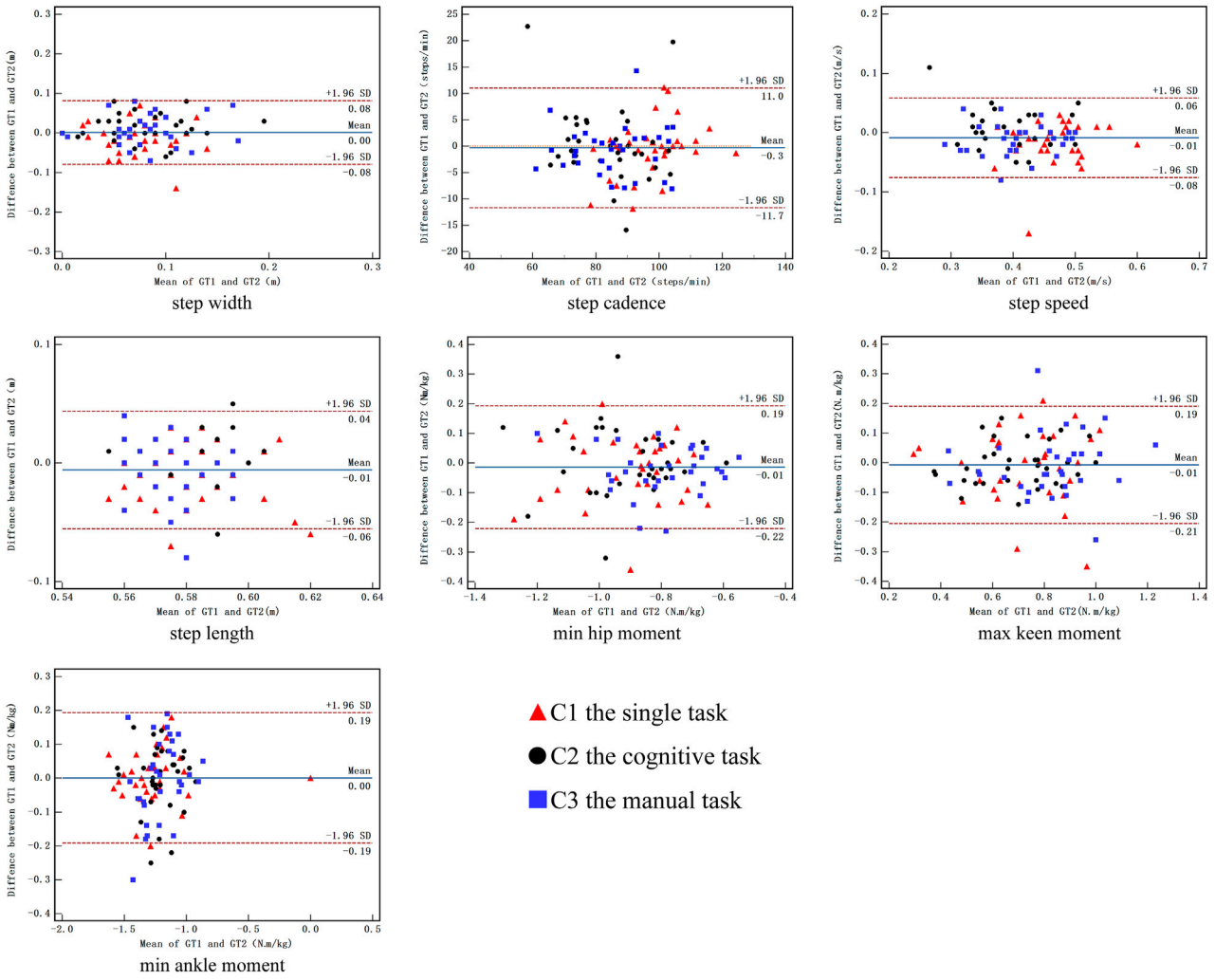

FIGURE 2
Bland-Altman plot of the difference between session 1 and session 2 against mean value during ascending stair for kinematic and kinetic parameters. C1, the single task; C2 the cognitive task; C3 the manual task.

0.873 in step speed, from 0.185 to 0.597 in step length, and from 0.500 to 0.660 in step width under the three conditions.

Table 4 shows that the interclass correlation coefficient values ranged from 0.133 to 0.963 in minimum hip moment, from 0.893 to 0.935 in maximum knee moment, and from 0.057 to 0.850 in minimum ankle moment under the three task conditions. The relationship values ranged from 0.075 to 0.930 in minimum hip moment, from 0.816 to 0.887 in maximum knee moment, and from 0.030 to 0.752 in minimum ankle moment under the three conditions.

Figure 3 shows the Bland—Altman plots of the kinematic and kinetic parameters during descending stairs. The numbers of dots outside the range of 95% CI were one dot (3.1%) in step width and step cadence in the single task, two dots (6.3%) in step width to four dots (12.5%) in step cadence in the cognitive task, and three dots (9.4%) in step width to one dot (3.1%) in step cadence in the manual task. The numbers of dots outside the range of 95% CI were one dot (3.1%) in step speed in the single task, two dots (6.3%) in step length in the cognitive task, and two dots (6.3%) in step speed to one dot (3.1%) in step length in the manual task. The numbers of dots outside the range of 95% CI were one dot (3.1%) in maximum knee moment in the single task, two dots (6.3%) in maximum knee moment in the cognitive task, and one dot (3.1%) in maximum knee moment to three dots (9.4%) in minimum hip moment in the manual task. The numbers of dots outside the range of 95% CI were one dot (3.1%) in minimum ankle moment in the single task and cognitive task and two dots (6.3%) in minimum ankle moment in the manual task.

4 Discussion

This study revealed the test—retest reliability of kinematic and kinetic parameters in the elderly during stair walking under single task, cognitive task, and manual task conditions (Flansbjer et al., 2005; Gorelick et al., 2009). Most of the test–retest reliability values for these parameters were good to excellent.

TABLE 3 The test-retest reliability of kinematic parameters under three conditions when descending stairs.

		Firs session	Secon session	ICC	95% CI	r	P
Single task	Step cadence (step/min)	109.1 ± 16.1	109.2 ± 16.6	0.956	0.909–0.978	0.916	<0.001
	Step speed (m/s)	0.56 ± 0.08	0.55 ± 0.09	0.878	0.750–0.940	0.785	<0.001
	Step length(m)	0.63 ± 0.02	0.63 ± 0.02	0.745	0.478–0.876	0.597	<0.001
	Step width(m)	0.07 ± 0.04	0.07 ± 0.04	0.795	0.580–0.900	0.660	<0.001
Cognitive task	Step cadence	90.4 ± 19.4	91.9 ± 19.8	0.944	0.886–0.973	0.895	<0.001
	Step speed (m/s)	0.48 ± 0.10	0.47 ± 0.10	0.932	0.861–0.967	0.873	<0.001
	Step length(m)	0.62 ± 0.02	0.62 ± 0.02	0.710	0.597–0.885	0.396	0.025
	Step width(m)	0.08 ± 0.03	0.08 ± 0.04	0.685	0.354–0.846	0.528	0.002
Manual task	Step cadence	91.4 ± 13.7	93.6 ± 15.4	0.925	0.846–0.963	0.866	<0.001
	Step speed (m/s)	0.47 ± 0.08	0.47 ± 0.08	0.779	0.548–0.892	0.639	<0.001
	Step length(m)	0.62 ± 0.02	0.62 ± 0.02	0.305	−4.25–0.661	0.185	0.310
	Step width(m)	0.06 ± 0.04	0.06 ± 0.03	0.661	0.306–0.835	0.500	0.004

TABLE 4 The test-retest reliability of kinetic parameters under three conditions when descending stairs. (N m/kg).

		First session	Secon session	ICC	95% CI	r	P
Single task	Min hip moment	0.54 ± 0.28	0.58 ± 0.30	0.963	0.924–0.982	0.930	<0.001
	Max knee moment	1.13 ± 0.25	1.08 ± 0.29	0.935	0.867–0.968	0.887	<0.001
	Min ankle moment	1.03 ± 0.19	1.07 ± 0.25	0.811	0.612–0.908	0.706	<0.001
Cognitive task	Min hip moment	0.55 ± 0.32	0.55 ± 0.28	0.958	0.913–0.979	0.928	<0.001
	Max knee moment	1.10 ± 0.24	1.14 ± 0.28	0.929	0.855–0.965	0.878	<0.001
	Min ankle moment	1.01 ± 0.25	1.04 ± 0.21	0.850	0.693–0.927	0.752	<0.001
Manual task	Min hip moment	0.50 ± 0.28	0.47 ± 0.20	0.133	−0.776–0.577	0.075	0.683
	Max knee moment	1.09 ± 0.23	1.11 ± 0.27	0.893	0.780–0.948	0.816	<0.001
	Min ankle moment	1.01 ± 0.19	0.98 ± 0.23	0.057	−0.933–0.539	0.030	0.871

ICC is a popular statistical method used to assess test–retest reliability (Chan and Pin, 2019) and is based on the calculation of the F-value from repeated-measures ANOVA. In this study, most of the test–retest reliability values of the kinematic and kinetic parameters were high in the single task, which supported the first hypothesis. The ICC value (0.511–0.979) demonstrated fair to excellent test–retest reliability in the single task, except for step length (Flansbjer et al., 2005). The finding was consistent with the study of Santos et al. (Santos et al., 2022), who reported that the agreement between manual stopwatches and photoelectric cells was excellent (ICCs between 0.92 and 0.97) when the distance was 5 m. Bland—Altman plot analysis is an important method used to detect systematic errors (Bland and Altman, 1999). The axis of ordinate was determined based on the differences between two measurements, and the axis of abscissa was determined based on the mean of two measurements. In some clinical studies, Bland—Altman analysis was used to evaluate the agreement between two testing sessions (Soulard et al., 2021). Reliability was assessed using the 95% limits of agreement and mean difference. As shown in Figures 2, 3, during single task, the zero values were included in the 95% CI, few dots were outside the range of 95% CI, and the mean difference was close to zero, indicating fair reproducibility (Dolatabadi et al., 2016). The good test–retest reliability of the kinematic and kinetic parameters during single-task stair walking might be attributed to several factors. First, the same testing control could help reduce measurement errors, including the same time interval between consecutive tests, the same testing commands, the same gait and postural control during stair walking, and the laboratory environment (Flansbjer et al., 2005). Second, three pre-test practices were provided to each participant before data collection to help them understand the procedure and minimize errors in learning (Sun et al., 2015). Finally, the test and retest sessions of reliability were scheduled a week apart, which could prevent learning and fatigue from affecting the reproducibility of the measurements.

In dual-task stair walking, the ICC and r values of the kinematic and kinetic parameters demonstrated fair to excellent test–retest reliability. The Bland–Altman plot also showed that the zero values were included

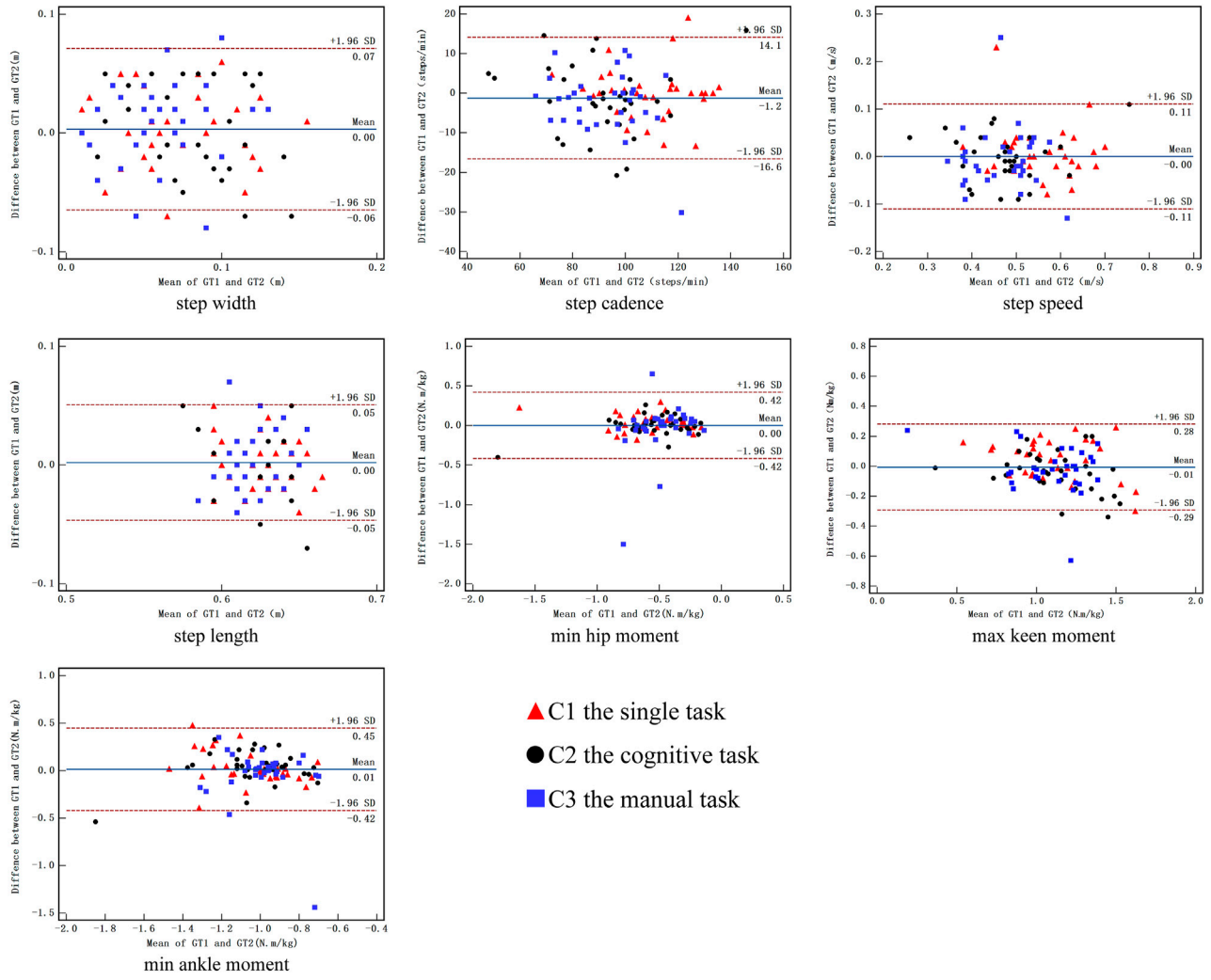

FIGURE 3
Bland-Altman plot of the difference between session 1 and session 2 against mean value during descending stair for kinematic and kinetic parameters. C1, the single task; C2 the cognitive task; C3 the manual task.

in the 95% CI, the mean difference was close to zero, and about 14 dots (87.5%) were inside the range of 95% CI during cognitive and manual stair walking. Our results showed that the test–retest reliability of the kinematic and kinetic parameters during dual-task stair walking was fair to excellent, which did not support the second hypothesis. However, previous studies were in line with our results. A recent study by Cho et al. showed that some level walking gait parameters (e.g., speed, stride length, cadence, etc.) while talking had good reliability (ICC = 0.69–0.90) under dual-task conditions (Cho et al., 2015). Tsang et al., (2019) also found that the test–retest reliability of the dual-task walking distance tests were excellent (ICC = 0.891–0.984). When confronted by two attention-demanding activities, humans prioritize one task over the other based on counterbalancing capabilities and cognitive availability (Yogev-Seligmann et al., 2012). The elderly might prefer to perform posture control of gait stability rather than deal with the "secondary" cognitive task during walking to minimize the loss of balance and maximize the avoidance of hazards (Shumway-Cook et al., 1997). When facing more complexed situations, such as stair walking, individuals tend to allocate attention resources for posture control to ensure safety (Wang et al., 2021). Therefore, the test–retest reliability of posture control during stair walking is high.

However, the ICC value ranged from 0.394 to 0.553, and the r value ranged from 0.203 to 0.246 of step length when ascending stairs under single- and dual-task conditions, indicating the poor test–retest reliability of step length. The ICC values were 0.133 in minimum hip moment and 0.057 in minimum ankle moment; the r values were 0.075 in minimum hip moment and 0.030 in minimum ankle moment in the manual task when descending stairs, which showed the poor test–retest reliability of both parameters. Our finding was inconsistent with the results of this study (Ryan et al., 2006); at all speeds, inter device reliability was excellent for the activPAL physical activity monitor ((ICC (2,1)≥0.99)) for both step number and cadence. The discrepancy might be related to the differences in gait complexion between stair and level walking. More challenging and complex tasks might require more attentional resources (Ruffieux et al., 2015). Stair walking, as the most hazardous aspect of gait, might be more challenging and complex and

requires greater attentional resources (Mian et al., 2007). According to capacity–sharing theory, cognitive resources are limited in capacity; when two tasks sharing common neural circuitry are performed at the same time, both of them are processed, but the processing ability between the tasks will decrease (Patel et al., 2014). When performing stair walking and dual tasks simultaneously (Wang et al., 2021), the poor test–retest reliability may have occurred due to the competition of limited attention resources, resulting in the decline of task performance in dual-task situation during stair walking (Ettwig and Bronkhorst, 2015). This study might provide reference for the selection of kinematic and kinetic parameters during dual-task stair walking.

This study has three limitations. First, the sample size was relatively small, and the test–retest reliability of stair walking should be used cautiously for clinical assessment. Second, all the participants were healthy elderly adults, and further studies should focus on the reliability of stair walking in frail elderly adults with a high fall risk. Finally, we only recruited people with a BMI of 18.5–25 Kg/m^2, so the result should be cautiously used for the elderly population.

5 Conclusion

The results obtained from this study show the good test-retest reliability of step cadence, step speed, and step width during single- and dual-task stair walking in the elderly, and the poor reliability of step length when ascending stairs. All the kinetic parameters, including min hip moment, max knee moment, and min ankle moment, had good test-retest reliability during single- and dual-task stair walking, but min hip moment and min ankle moment had poor reliability during manual-task descending stairs. These results may help researchers in the assessment of biomechanics of dual-task stair walking in the elderly and to interpret the effect of interventions in this population.

Data availability statement

The original contributions presented in the study are included in the article/supplementary material, further inquiries can be directed to the corresponding author.

Ethics statement

The studies involving human participants were reviewed and approved by the Shandong Provincial Natural Science Foundation (No. ZR2020QC091). The patients/participants provided their written informed consent to participate in this study. Written informed consent was obtained from the individual(s) for the publication of any potentially identifiable images or data included in this article.

Author contributions

YL and WS are responsible for drafting this protocol and participating in the conception, study design, assessments, data interpretation, and writing and submission of the manuscript. JW and CZ are responsible for the study design and revision of the manuscript. QS and CZ are responsible for the study design. JW and NY are responsible for the data management and monitoring of this study. All authors have read the manuscript and approved the publication of this protocol.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Effects of various hyperopia intervention levels on male college students' gait kinematics

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Background: Hyperopia is a common blurred vision phenomenon that affects postural control in gait; however, current research has focused on the alteration and correction of hyperopia's physiological characteristics, ignoring the effect of hyperopia on gait kinematic characteristics. The effect of hyperopia on the basic form of movement walking is a worthy concern.

Objective: To investigate the gait kinematic characteristics of male college students with varying degrees of visual acuity (normal vision, hyperopia 150°, and hyperopia 450°), as well as to provide a theoretical foundation for the effect of visual acuity on gait and fall risk reduction.

Methods: Twenty-two male college students with normal visual acuity were chosen. Their vision was tested using a standard visual acuity logarithm table at normal and with 150° and 450° concave lenses. Gait kinematic data were collected under normal vision and hyperopic conditions using the PN3 Pro advanced inertial motion capture system and Axis Studio application program.

Results and conclusion: 1. The change of center of gravity in Pre-double support was smaller than normal vision; Late-single support and Late-swing was larger than normal vision; 2. The percentage of the double-leg support decreased; the percentage of the single-leg support and the Late-swing increased; 3. For the joints' range of motion, Trunk flexion and extension range of motion in Pre-single support, Late-double support and Pre-swing smaller than normal visual acuity, and Late-swing larger than normal; hip internal abduction and adduction and internal and external rotation are larger than normal vision in Late-single support; knee and ankle in abduction and adduction direction are larger than normal vision in the swing stage; hip flexion and extension, internal external rotation are larger than normal vision in the swing stage. Hyperopic interventions have an impact on the kinematic characteristics of gait in male college students, mainly in terms of altered balance, increased instability, increased difficulty in maintaining trunk stability, and increased risk of injury.

KEYWORDS

hyperopia, gait, kinematics, center of gravity, joint angle, gait cycle

1 Introduction

Walking is a fundamental mode of locomotion in humans, a physiological activity controlled by the nervous system and involving the coordinated action of muscles, bones, and joints throughout the body (Morelli and Morelli, 2021; Chen et al., 2022). Vision, vestibular, and proprioception are all important factors in gait, and vision, as the primary sensory input, can accurately respond to the body's relative position to the external environment, regulate gait postural control, and prevent falls (Morelli and Morelli, 2021). Postural reflexes and anticipatory postural regulation are reduced when there is visual impairment, affecting the body's postural control. Hyperopia is a common visual abnormality that affects the input of visual information in gait. However, the hyperopia people will only wear glasses in the near vision situation such as reading books and newspapers, but the abnormal posture control caused by hyperopia exists all the time, so the posture control during the gait of the hyperopia people is worth studying (Chinese Journal of Rehabilitation Medicine, 1993).

In the study of gait in sports biomechanics, kinematic data are often collected using kinematic data acquisition by infrared capture systems and inertial motion capture systems, kinetics are analyzed using dynamometers and plantar pressure, and muscle force is analyzed using electromyography (Brodoehl et al., 2015; Wiesinger et al., 2022). Based on the fact that changes in kinetic and EMG-related data are important causes of alterations in gait kinematic characteristics, the external appearance of postural control imbalances can be found in kinematic analysis. Kinematic analysis is a scientific method to study the changes in the temporal and spatial movement patterns of the limbs during walking. The kinematic changes can be reacted to the center of gravity (Lehman et al., 2008), the percentage of gait cycle (Yang and Chen, 2019; Ma et al., 2021) and joint motion angles (You et al., 2003; Dai and Li, 2016; Matchar et al., 2017; Wang et al., 2018; Liu et al., 2022). The more stable the center of gravity and the relatively more the proportion of double support, the better the postural control.

Visual impairment (Morelli and Morelli, 2021) and visual fatigue (Davids, 2014) have been evidenced to affect balance and produce falls, but there is insufficient evidence to demonstrate the effect of vision, especially hyperopia, on postural control in gait. The purpose of this study was to investigate the effect that different levels of hyperopia would have on the kinematic characteristics of gait. The hypothesis of this study is that changes in center of gravity, the percentage of gait cycle and joint motion angles will take effect after a hyperopic intervention.

2 Subjects and methods

2.1 Designs

Experiment datas on gait kinematics for male college students were used repeated measures of one-way ANOVA under normal vision and hyperopia conditions of 150° and 450°.

2.2 Time and location

The experiment will last 20 days and begin in July 2022. The test will take place at Zunyi Medical University's Experimental Center of Sports Rehabilitation.

2.3 Participants

Thirty male college students with normal vision from Zunyi Medical University were recruited. After eligibility assessment, a total of 22 participants met the experimental requirements, the participants' average age is 20.82 ± 1.40 years, their average height is 174.86 ± 3.27 cm, their average weight is 65.48 ± 9.48 kg, and their average BMI is 21.32 ± 2.54 kg/m^2. Before the trial, all participants provided written informed consent and understood the experimental process and purpose. This study was reviewed and approved by the Ethics Committee of the Zunyi Medicai University.

Criteria for Inclusion: 1. Normal vision in both eyes (visual acuity of both naked eyes 5.0); 2. No recent history of eye or vision correction surgery; and 3. No impairment of walking or motor function.

Criteria for Exclusion: 1. Exclusion Poor visual acuity, defined as visual acuity of less than 5.0 in either eye; 2. The presence of motor impairment; 3. The inability to finish the exam on one's own exclusion factors.

2.4 Methods

2.4.1 Data collection for visual acuity

The visual acuity of the participants is collected by professional testers when they are in normal vision and wearing 150° and 450° concave lenses. The environment of the place is required to be clean, neat and quiet; the size of the place and the light meet the standard, the distance is up to the standard, the illumination is uniform, constant, non-reflective and non-dazzling, and direct sunlight is prohibited. The visual acuity test was conducted using a standard logarithmic visual acuity table, which consisted of 12 rows of "E" with different sizes and openings in different directions, and measured visual acuity in the range of 4.0–5.2; each row was labeled with a number, the visual acuity table was hung at a height such that most of the measures had a horizontal line of sight at the position of 5.0 rows of visual. The participants were in a quiet state 10 minutes before the test and could not perform a strenuous exercise; during the test, the participants were kept upright and was randomly selected normal vision, 150° or 450° hyperopic intervention by the tester to test the visual acuity. The right eye test is as follows: the participants pointed out the direction of each viewpoint in turn, starting from the largest viewpoint. The examination was completed when the participant made one error in each row from 4.0 to 4.5, or two errors in each row from 4.6 to 5.0, or three errors in each row from 5.1 to 5.3. The visual acuity represented by the previous row is the result of the participant's visual acuity test. The results were recorded according to the five-point recording method. To guarantee the accuracy of the data, 5% of the participant were chosen for evaluation after the visual acuity test.

Grading of visual acuity (Tong et al., 2000; Vivekanand et al., 2021): Mild hyperopia was defined as 300° of hyperopia, moderate hyperopia as 300°–600° of hyperopia and Severe hyperopia as >600° of hyperopia; intermediate values of 150° and 450° of low and moderate hyperopia, respectively, were chosen. To simulate hyperopia, 150° and 450° concave lens glasses were used: 150° concave lens glasses were used for mild hyperopia, and 450° concave lens glasses were used for moderate hyperopia. The test was not performed because high hyperopia is very blurry.

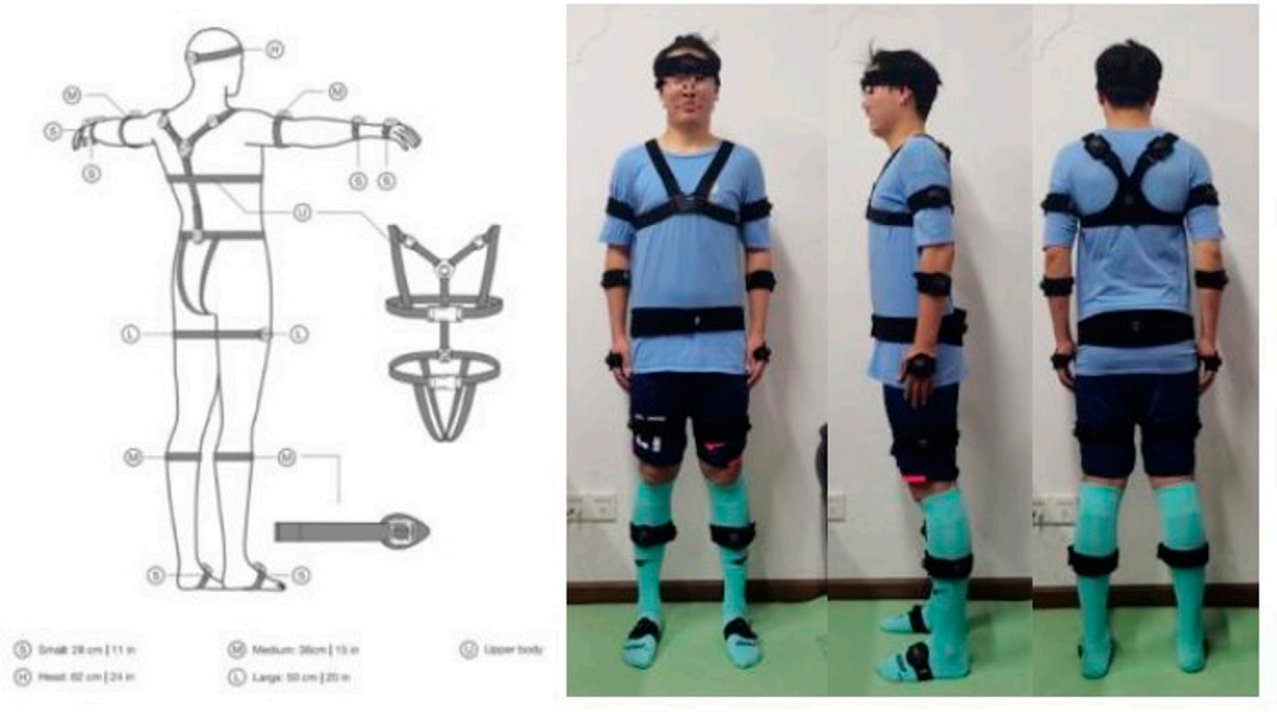

FIGURE 1
Wearing sensors.

2.4.2 Data collection for gait

2.4.2.1 Experimental equipment

The Perception Neuron 3 Pro advanced inertial motion capture system was used to collect gait kinematic data for this study. Morphological data such as head length, neck length, torso length, shoulder width, femur width, upper arm length, forearm length, hand length, thigh length, calf length, foot length, heel height, arm span, and so on were measured and recorded by professional testers and imported into the skeletal size database to create a personalized model of the participant. Wearing sensors: 16 sensors were placed at the midpoint of the forehead, the upper third of the scapulae, the forearm, the lateral midpoint of the thigh, anterior side of calf, the fourth lumbar vertebra, the dorsum of the foot and the dorsum of the hand (as shown in Figure 1). After the participant has finished wearing the device, the software is opened for warm-up, the sensors are connected, and the participant performs posture calibration using the A-T-B-P action (as shown in Figure 2): A-posture is when the body is upright, the arms are down, and the palms of the hands face the body. Adjust the space between your feet to be the same width as your hips, with both feet facing forward. The T-pose is based on the A-pose, with the arms raised sideways to parallel to the horizontal, while the palms face down. The B-Posture is based on the A-Posture with the shoulders slightly extended forward and the elbows flexed at 90°, while the palms are placed against each other in the middle of the chest; and the elbows, wrists and middle fingers are in a line, while the thumbs and other four fingers are at an angle of about 45°. P-position is to straighten the thumb and gently pinch the index finger, while requiring the thumb to be located in a straight line from the heel to the tip as much as possible, with the index finger naturally bent to touch the tip of the thumb and the remaining three fingers naturally bent. When the calibration is finished, a personalized model is created and ready for testing.

2.4.2.2 Gait data collection process

Participants wore 150°, 450° concave lenses, or normal without glasses and walked barefoot independently in a sensor sensing area of 10m*3m, with the requirements of the test being uniform speed and smooth walking. If there is external interference, a sensor signal that is outside the test range, or an abnormal sensor signal during the experiment, the test is considered a failure and must be re-tested. Three times successful gait kinematic data collections for normal vision and hyperopia 150° and 450° vision were completed.

2.4.2.3 Data processing

The moment of landing and the moment of leaving the ground is the key point to divide the gait cycle, the gait cycle can be defined as one side of the feet following the ground to the side of the heel landing again, according to the key point, we transfer the sensor data to Axis Studio software, wthere the test video is simulated at the 3D level based on kinematic data and divided a gait cycle into: Pre-double support, Pre-single support, Late-single support, Late-double support, Pre-swing and Late-swing. Pre-double support: left heel pointing-right toe off the ground; Pre-single support: right toe off the ground - right heel passing the left foot; Late-single support: right heel passing the left foot-right heel landing; Late-double support: right heel landing-left toe off the ground; Pre-swing: Left toe off the ground-Left heel passing right heel. Late-swing: Left heel passing right heel-Left heel pointing to the ground. (Yang and Chen, 2019).

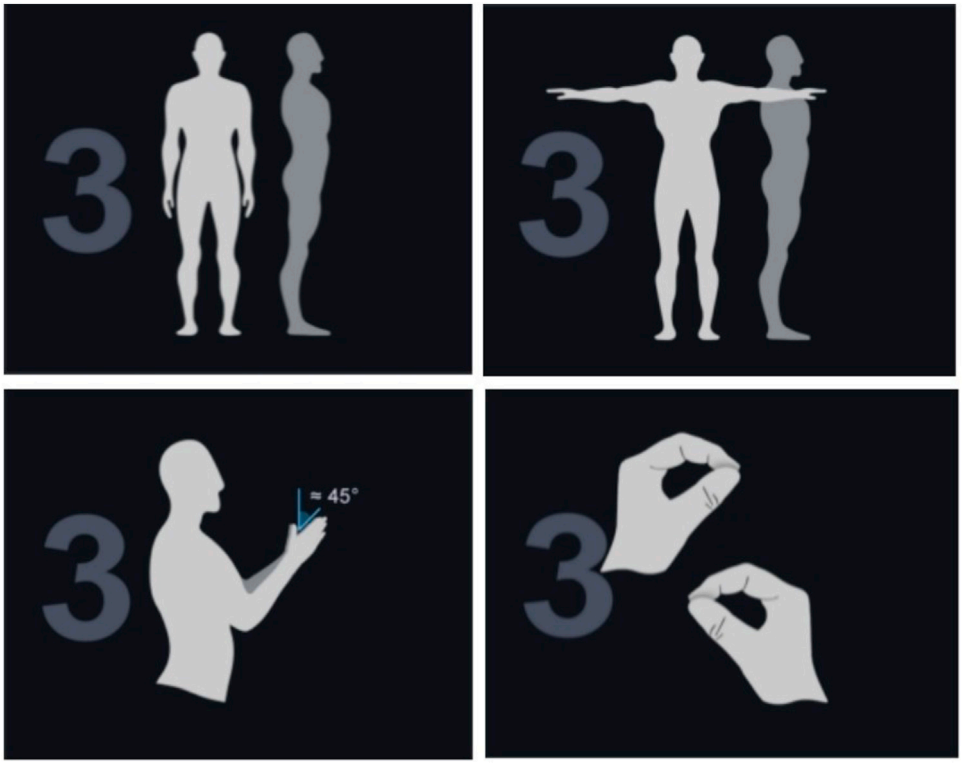

FIGURE 2
A-T-B-P posture.

The following are the main collection indicators: the range of vertical center of gravity change in each phase of the gait cycle, percentage of each phase of the gait cycle, the joint motion angle of the trunk and the lower limbs of the hip, knee and ankle around the frontal, sagittal and vertical axes.

2.5 Statistical analysis

Statistical analysis of indicators was performed using SPSS 24.0 software, and the data were described by mean ± standard deviation. One-way ANOVA with repeated measures was used for comparison between groups, and LSD multiple comparisons were used for *post-hoc* tests, with the significance level taken as $\alpha = 0.05$.

3 Results

3.1 Visual acuity test results

According to Table 1, Normal vision tests for male university students, all with visual acuity values above 5.0, with a 150° concave lens: left eye vision is 5.11–5.19, right eye vision is 5.10–5.20 and binocular vision is 5.17–5.22; with a 450° concave lens: left eye vision is 4.58–4.91, right eye vision is 4.57–4.89 and binocular vision is 4.52–4.84.

3.2 Change of center of gravity

Figure 3 demonstrates that for male college students with hyperopia of 150° and 450°, the range of change in center of gravity in Pre-double support was significantly smaller than normal vision ($p < 0.05$); the range of change in center of gravity in Late-single support and Late-swing was significantly larger than normal vision ($p < 0.05$); and only the hyperopia of 450°, all gait cycle's time lager than normal visual ($p < 0.05$).

3.3 Percentage of gait cycle

As shown in Figure 4, in the hyperopic 150° and 450° conditions the percentage of the gait cycle in Late-single support and Late-swing was significantly larger than normal vision ($p < 0.05$); in the hyperopic 450° condition, the percentage of the gait cycle in Late-double support and the all gait cycle's time was significantly less than normal vision ($p < 0.05$).

3.4 The trunk's range of motion

Figure 5 shows that in the 150° and 450° hyperopia in the Late-double support of the trunk lateral flexion direction, in the Pre-swing of the trunk flexion and extension direction for male college students were significantly smaller than normal visual acuity ($p <$

TABLE 1 The basic information of participants' visual

		Visual acuity	Lower limit	Upper limit
Vision in the left eye	Normal	5.13±0.09	5.09	5.17
	hyperopia 150°	5.15±0.09	5.11	5.19
	hyperopia 450°	4.75±0.38	4.58	4.91
Vision in the right eye	Normal	5.15±0.10	5.11	5.19
	hyperopia 150°	5.15±0.11	5.10	5.20
	hyperopia 450°	4.73±0.37	4.57	4.89
Binocular vision	Normal	5.22±0.08	5.18	5.25
	hyperopia 150°	5.20±0.07	5.17	5.22
	hyperopia 450°	4.68±0.36	4.52	4.84

Table Note: Hyperopia 150° is the visual acuity measured by wearing a 150° concave lens, and hyperopia 450° is the visual acuity measured by wearing a 450° concave lens.

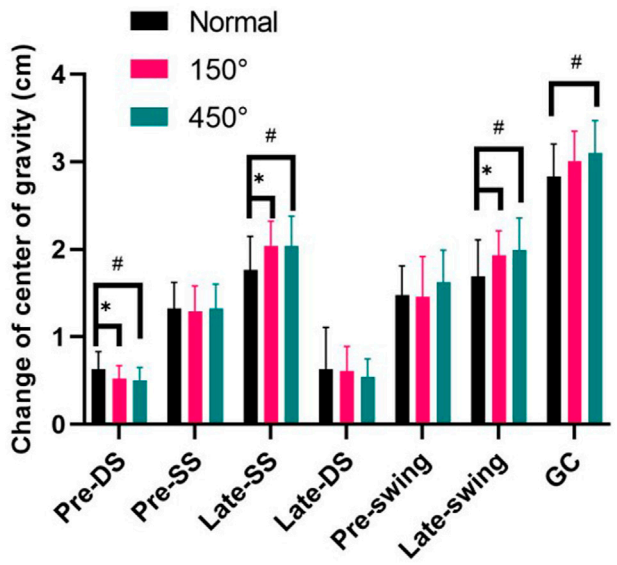

FIGURE 3
Change of center of gravity.

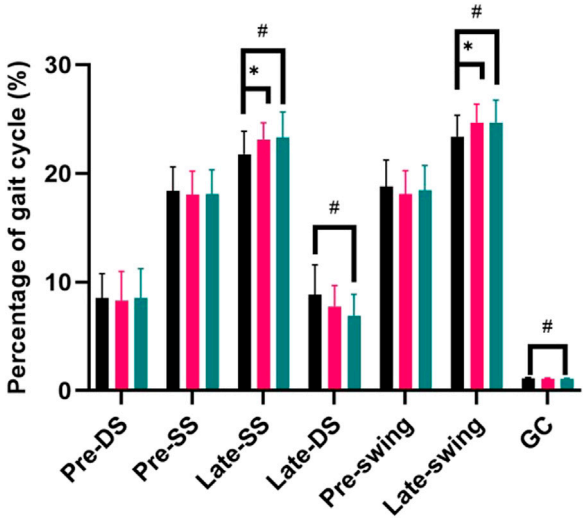

FIGURE 4
Percentage of giat cycle.

0.05). In the 150° hyperopia, in Pre-double support the trunk joint movement of the lateral flexion and rotation were significantly smaller than normal visual acuity ($p < 0.05$), and in the 450° hyperopia, in Pre-double support, Pre-single support and Late-single support, the trunk joint movement of the flexion and extension were significantly smaller than normal visual acuity ($p < 0.05$).

3.5 The lower limb joints range of motion

The range of variation was larger in the Late-single support and Late-swing, as shown in Figure 6. In 150° and 450° hyperopia, in the Pre-single support the hip joint motion of the internal abduction and adduction were significantly smaller than normal visual acuity in the Late-single support the hip joint motion of the internal abduction and adduction, internal rotation and external rotation, and ankle joint motion of the flexion and extension were significantly larger than normal visual acuity; in the Pre-swing the hip joint motion of the internal rotation and external rotation and in the Late-swing the hip joint motion of the internal rotation and external rotation and flexion and extension were significantly larger than normal visual acuity ($p < 0.05$). In 150° hyperopia the knee joint motion of the internal abduction and adduction in the Late-double support was significantly larger than normal visual acuity; and the knee joint motion of the internal abduction and adduction was significantly larger than normal visual acuity in the Late-swing and Late-single support ($p < 0.05$). In 450° hyperopia, in the Pre-swing the knee motion of the internal rotation and external rotation was significant differences with normal visual

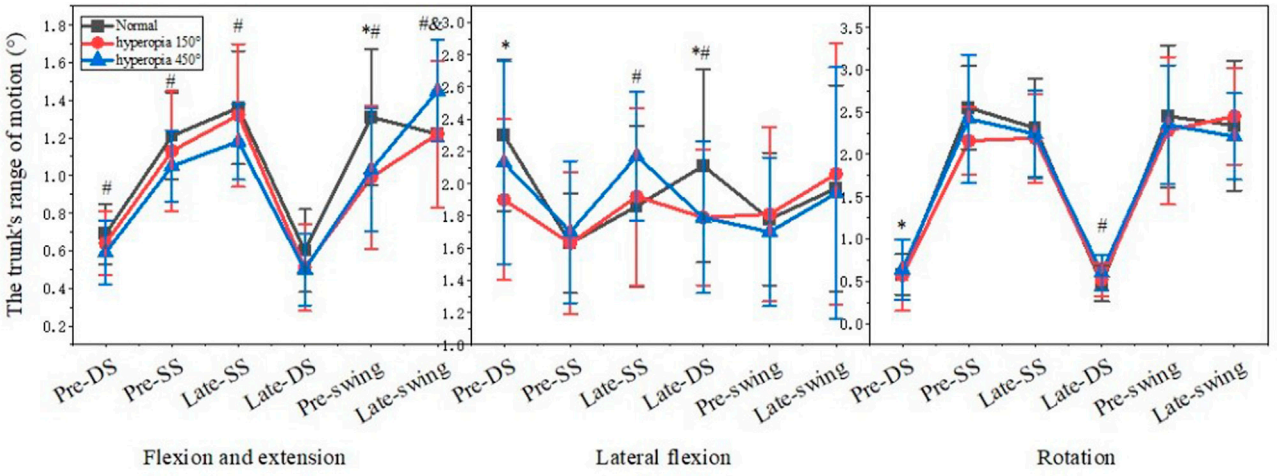

FIGURE 5
The trunk's range of motion.

acuity; in the Late-swing the ankle joint motion of the internal abduction-adduction, was significantly larger than normal visual acuity ($p < 0.05$).

4 Discussion

Hyperopia is a condition in which the eye at rest forms a focal point on the back of the retina of parallel light. It is caused by a physiological or pathological shortening of the anterior posterior axis of the eye, a reduction in the surface curvature of the refractive body, and a weakening of the lens's refractive power. Light pollution is becoming more serious as society develops, people's vision is gradually deteriorating, and hyperopia is also trending towards youth. Hyperopia affects postural control and muscle preactivation of the body's movement between travels; for example, human gait balance is primarily dependent on visual regulation. In our study, using 150° and 450° concave lens glasses, investigated the effects of different degrees of hyperopia on the kinematic characteristics of male college students' gait. During the visual acuity test, the 450° hyperopia results were significantly smaller than normal visual acuity and 150° hyperopia, and there was no significant difference between the normal visual acuity and in the 150° hyperopia, but the participants' gait kinematic characteristics changed at 150° hyperopia, which may be due to the fact that the participants wore 150° concave lenses to test their visual acuity by squinting, they adapted to the change in visual acuity, and the visual acuity test is a relatively static process, whereas walking is a dynamic process that is participant to changes in the surrounding environment and requires more adjustment, so the visual acuity test results did not differ significantly from normal visual acuity at 150° of hyperopia, but the kinematic characteristics of the gait cycle did.

The vertical center of gravity trajectory of the body shows a sinusoidal trajectory motion of about 5°, while the pelvic trajectory is a forward and backward rotational motion of about 8° during the gait cycle. Non-physiological changes in the center of gravity's trajectory may indicate an increased risk of movement (Bolink et al., 2012; Eckel et al., 2012). For example, a small range of body center of gravity changes makes the body more stable and less likely to fall when walking, whereas an increased range of center of gravity changes may impair balance function (Lefeber et al., 2020; Moon et al., 2022). The results showed that the change in center of gravity during the Late-double support in male college students with hyperopia vision was smaller than normal vision, which may be due to the abnormal visual input under the hyperopia condition, which cannot greater identify the spatial position of the body and relies on reducing the change in center of gravity to maintain body stability; however, the range of change in center of gravity during the single-leg support was larger than normal vision. Because the body is unable to regulate the change in center of gravity, there is a risk of falling (Zhao and Zhou, 2003; Liu et al., 2022). Therefore, balance training should be increased for the hyperopia, especially the elderly, to increase the stability of the hyperopia elderly during walking and reduce the occurrence of falls.

The stability of the human body during the double-support period is larger than that of the single-leg support throughout the gait cycle (Duclos et al., 2012). In double-leg support, both feet are on the ground, the center of gravity is offset to the both feet, and the body weight is borne by both feet; in single-support, one foot is off the ground, the center of gravity is offset to the supporting one leg, and the body weight is borne by one lower limb; and the support area of the center of gravity is reduced compared to that of double-support, so the more the proportion of the double-support, the more stable the body (Liu et al., 2022). In our study, in the hyperopic condition the percentage of the double-leg support decreased; the percentage the single-leg support and the Late-swing increased, which are consistent with a previous study, which found that the proportion of single-leg support time in obese children was higher than in normal-weight children (D'Hondt et al., 2011). Therefore, it can be inferred that hyperopia leads to a certain degree of change in

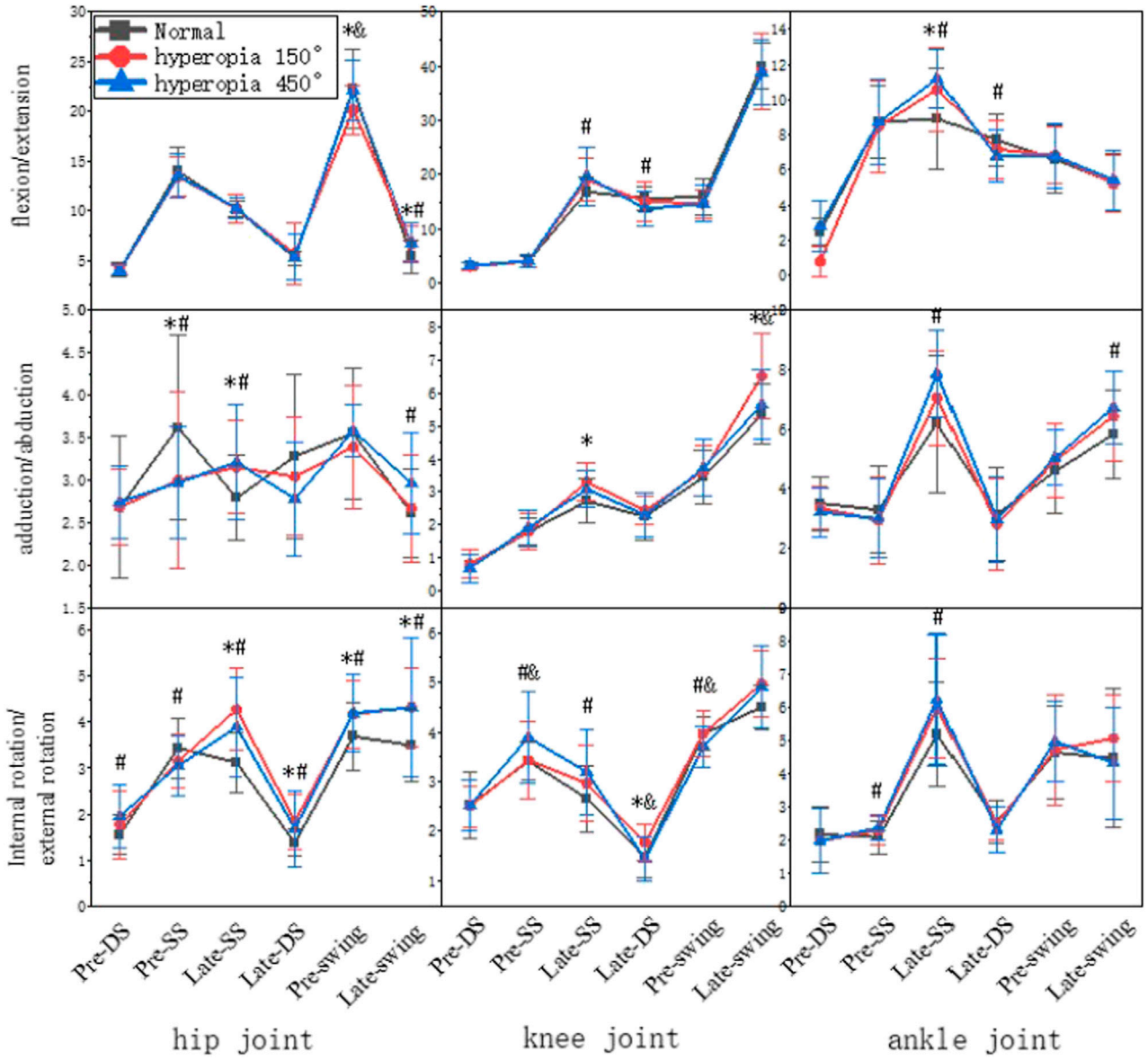

FIGURE 6
The lower limb joint's range of motion.

the percentage of gait cycle, resulting in an increase in instability time and a decrease in stability time, which affects the stability of the whole gait cycle. Therefore, stability training for hyperopic people is also very important, especially in the single-leg support state.

The trunk plays a pivotal role in gait, and its non-physiological forward flexion and extension as well as lateral flexion and rotation can cause movement tension and affect walking stability; trunk posture control is closely related to the angle of lower limb joint movements, and changes in trunk posture cause changes in the angle of lower limb joint movements, and lower limb joint movements also regulate changes in trunk posture (Lang and Ji, 2003). Studies have shown that anterior trunk flexion causes forward head flexion and tends to cause anterior pelvic movement, limiting forward and backward hip rotation around the vertical axis and reducing stride length; posterior trunk extension tilts the head back, leading to a longer support phase (Haruyama et al., 2019). In addition, the backward tilt angle of the trunk is too large to cause the body to form an anti-support state; the forward pitch angle is too large to limit the flexibility of the hip joint; the left and right swing amplitude is large to generate excess horizontal component forces, which tend to affect the ability of visual and vestibular sensory regulation of posture, all of which are detrimental to the stability of human walking. In this study, the trunk flexion and extension angle in the Late-swing under the hyperopic 450° condition was larger than normal, at same time, the hip inversion and abduction angle was significantly larger than normal, and this phenomenon may be due to the fact that the hip and trunk are adjacent to each other, and the increase in trunk flexion and extension angle may be related to the

hip angle and the proportion of each phase in the gait cycle, and it is mainly the rectus femoris muscle in the Late-swing that makes the lower leg swing forward and maintains the stability of the lower leg in the Late-swing, and it is also one of the main force-generating muscles in hip flexion (Xiong et al., 2022). Thus, it was confirmed that trunk postural control and hip joint movement angle are closely related and that the transmission, analysis and integration of sensory information related to trunk control is reduced in participants, which affects trunk control (Petersen et al., 2022). Therefore, this study suggests that people with hyperopia should strengthen the training of core muscles to increase the postural control of the body in order to reduce the probability of falling.

Postural control is primarily regulated by sensory inputs such as visual, kinesthetic, and vestibular sensations. Changes in the structure and function of the visual system weaken visual sensory information input and make it more difficult for the body to identify spatial position and motion feedback while walking, which eventually leads to imbalance in postural control and falls, so the body will regulate by changing the relative position of joints or joint range of motion to avoid danger. In our study, the three lower limb joints are altered to varying degrees of joint range at various stages, and the stages with the most differences in joint changes are the Late-single support and Late-swing, which may be one of the reasons why most falls occur in Late-single support and Late-swing for example, studies have shown that falls in older adults occur in greater numbers in Late-single support and at the moment of the landing. Injuries in athletes occur more frequently during the swing period, and the risk of injury in both the single-leg support and the swing phase is higher than in the other phases (Duclos et al., 2012; Ji et al., 2015; Niu et al., 2020). Based on the foregoing, it can be deduced that the phases with the greatest variation in the range of motion of the participants' joints under hyperopic conditions are the Late-single support and Late-swing, both of which are injury-prone.

Joint coordination is essential for maintaining gait stability (Ippersiel et al., 2021; Tang et al., 2022). Visual input is impaired, walking is unstable, and the body regulates mechanoreceptor input-output by inducing plastic changes in the central nervous system, strengthening information transmission pathways, enhancing proprioception, and repeatedly adjusting and controlling body center of gravity and limb position (Ribeiro and Oliveira, 2007; Tang et al., 2022). According to the findings of this study, the lower limb joint motion angle of male college students is increasing, indicating that the non-physiological increase in the lower limb motion angle of gait in the long-term hyperopia population can affect the trajectory of the center of gravity and predispose to chronic wear and tear of the lower limb joints, the increased angle of motion in the direction of knee flexion and extension increases the load on the anterior cruciate ligament, which can easily cause knee injuries. Furthermore, the rising trend of lower limb joint motion angle occurs mostly during the single support phase and swing phase, when the human body has a small support area and the range of change of center of gravity increases, increasing the risk of falling and injuries. Therefore, people with hyperopia should reduce the risk of injury through joint stability and coordination training.

4.1 Limitations

There are some limitations in this study: 1. Walking is a coherent movement done by the whole body in coordination, and the influence of vision on trunk, upper limb and lower limb coordination should be considered comprehensively. 2. Walking speed is not strictly defined in this experiment, which may have some influence on the experimental results. 3. For the study of center of gravity, this experiment selected the change of center of gravity in vertical direction, therefore in future studies, the change of center of gravity in up and down, left and right, and front and back should be selected for a comprehensive study. 4. In this study, we used a concave lens to intervene in a normal population, and even though we gave the participants enough adaptation time, there is still a possibility that it may not be able to fully simulate the hyperopic state, and there is a gap between it and the truly hyperopic population.

5 Conclusion

Hyperopic interventions have an impact on the kinematic characteristics of gait in male college students, mainly in terms of altered balance, increased instability, increased difficulty in maintaining trunk stability, and increased risk of injury. Therefore, it is recommended to increase gait training and increase postural control in walking for hyperopic older adults to reduce falls and injuries and improve quality of life in old age.

Data availability statement

The original contributions presented in the study are included in the article/Supplementary Material, further inquiries can be directed to the corresponding author.

Ethics statement

The studies involving human participants were reviewed and approved by Ethics Committee of Zunyi Medical University. The patients/participants provided their written informed consent to participate in this study.

Author contributions

ZZ and AX were responsible for experimental design and paper writing, ZZ, XZ and ZJ were jointly responsible for the implementation of the research protocol, data collection, analysis and processing, and WH was responsible for theoretical guidance and paper revision as well as guidance on data processing. All authors contributed to the article and approved the submitted version.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Effects of peripheral electromagnetic stimulation after an eccentric exercise-induced delayed-onset muscle soreness protocol in professional soccer players: a randomized controlled trial

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Introduction: To examine the effects of peripheral electromagnetic stimulation in male professional soccer players on markers of Delayed Onset Muscular Soreness (DOMS), induced by a protocol of exercise (60 min of eccentric and plyometric).

Methods: A randomized controlled trial with fourty-five professional soccer players aged 22.33 ± 4.82 years participated in the study. Twenty-three participants were assigned to the experimental group with peripheral electromagnetic stimulation (5 stimulations of 5 s at 100 HZ with 55 s of rest for a total of 5 min of treatment) and the remaining 22 participants were assigned to the control group. Pain pressure threshold (PPT) of the vastus medialis, the Visual Analogue Scale-Fatigue (VAS-F), half squat (HS) test and the maximum voluntary contraction of the quadriceps were assessed. All evaluations were performed before and after 1 h of the eccentric exercise induced DOMS, as well as at post 24–48, and 72 h.

Results: Group-by-time interaction was observed in PPT of the vastus medialis (p = 0.040) with a medium effect size (η^2 p = 0.069). From 48 to 72 h the experimental group showed an increase of PPT compared to control group (p = 0.015). There was no group-by-time interaction for HS, quadriceps strength and VAS-F ($p > 0.05$).

Discussion: Peripheral electromagnetic stimulation in male professional soccer players did not produce significant improvements in the power and strength of the lower limbs but decreased the peripheral sensitization of the vastus medialis after eccentric exercise protocol.

Clinical Trial Registration: https://www.anzctr.org.au/Trial/Registration/TrialReview.aspx?id=384050&isReview=true, Identifier: ACTRN12622000841774.

KEYWORDS
soccer, football, after-treatment, muscle soreness, rehabilitation

1 Introduction

Over the years, several investigations have tried to demonstrate the mechanism of appearance of the Delayed Onset Muscle Soreness (DOMS) (Sonkodi et al., 2020). DOMS can lead to a loss of joint amplitude of up to 50%, as well as a loss of muscle strength. DOMS is a common condition characterized by muscle pain, stiffness, and tenderness that typically develops 24–48 h after strenuous exercise or physical activity (Hickey et al., 2014; Hayashi et al., 2017). DOMS is caused by microscopic damage to the muscle fibers, which triggers an inflammatory response in the body. Athletes also report pain, that is, related to muscle fibers micro-damage. Some of these symptoms are related to inflammatory mechanisms. The most common is that DOMS appears when facing a return-to-play or an unusual exercise (Hickey et al., 2014; Hayashi et al., 2017). A current theory hypothesizes that the mechanism of DOMS is axonal compression. The peripheral nerves that take place in the neuromuscular spindle suffer micro-injuries, as well as a tunnel effect due to the compression generated by the muscles (Stifani, 2014; Radovanovic et al., 2015). The acute compression that could cause axonopathy may be due to the excitotoxicity of glutamate, which occurs in muscle fibers (Sonkodi, 2021). This new theory is explained in two phases: first, the damage to the neuromuscular spindle occurs, and then, there is a protective response-reflex that gives rise to symptoms such as stiffness and decreased strength (Pinho-Ribeiro et al., 2017). Clinicians try to find strategies to improve the symptoms of DOMS, to allow the return to training as soon as possible and without risks for the athletes. However, there is not a common consensus in the scientific community (Hohenauer et al., 2015; Davis et al., 2020).

In this line, peripheral electromagnetic stimulation is a non-invasive method to administer, through a rapid pulse of high-intensity electricity which causes a magnetic field in the periphery of the body, based on Faraday's laws and experiments. Peripheral electromagnetic stimulation is a painless therapy that uses low-frequency electromagnetic waves to treat musculoskeletal conditions. Peripheral electromagnetic stimulation has been used to relieve pain, reduce inflammation, improve blood circulation, and accelerate tissue healing. This therapy has been shown to have a positive impact on various musculoskeletal conditions, including back pain, osteoarthritis, tendinitis, and postoperative pain. In this article, we will discuss the effects of peripheral electromagnetic stimulation in musculoskeletal treatment (Barker, 1991). Peripheral electromagnetic stimulation works by delivering a low-frequency electromagnetic field to the affected area of the body. This field penetrates the skin and tissues, and interacts with the cells, tissues, and nerves in the area. The electromagnetic waves cause the cells to vibrate, which in turn increases blood flow and oxygenation to the area. This increased blood flow helps to reduce inflammation and swelling and promotes tissue healing. The electromagnetic waves also stimulate the production of endorphins, which are natural painkillers produced by the body. This helps to reduce pain and improve overall comfort. One of the primary benefits of PEMS is pain relief. It has shown that PEMS can effectively reduce pain in patients with musculoskeletal conditions (Barker, 1991). The stimulation produced at the neural level is similar to conventional transcutaneous electrical nerve stimulation (TENS) (Barker, 1991), but the transcutaneous magnetic stimulation allows a deeper penetration (Kanjanapanang and Chang, 2022). In fact, peripheral magnetic stimulation triggers massive proprioceptive stimuli when applied to muscles through two pathways: direct activation of sensorimotor nerve fibers. and indirect activation of mechanoreceptors in the muscle fiber (Sdrulla et al., 2015). In stroke patients, evidence shows an increase in regional cerebral blood flow of the premotor cortex and parietal areas in the injured hemisphere after applying peripheral magnetic stimulation to the paretic muscles (Meesen et al., 2011; Beaulieu and Schneider, 2015).

Peripheral electromagnetic stimulation tends to improve recovery through the new sensory input that it manages to generate. Said stimulation acts on the peripheral nerve, in our case the femoral nerve, being able to counteract the nociceptive signal that comes out of the axonal endings found in the muscle. Peripheral electromagnetic stimulation has an effect on the A beta fibers that are responsible for acute pain signals. It also achieves an improvement in the response, through a maximum recruitment of fibers with minimal skin stimulation, that is, involved in the process of an increase in strength and with it, in the improvement of recovery (Asao et al., 2022; Jiang et al., 2022).

Based on these findings and in order to improve the symptoms of muscle soreness, this study aimed to observe the effects of peripheral electromagnetic stimulation on performance and recovery in young professional soccer players suffering after an eccentric exercise-induced delayed-onset muscle soreness protocol. We hypothesized that the peripheral electromagnetic stimulation have positive effects in pressure pain threshold values of the vastus medialis, perception of fatigue during half squat, half squat speed and quadriceps peak force.

2 Methods

2.1 Study design

A double-blind randomized pilot study was carried out with professional soccer players, following the ethical principles of the Declaration of Helsinki. This study was elaborated following the Consolidated Standards of Reporting Trials (CONSORT) and was approved by the Research Ethics Committee of the (reference number: CIPI: 22.095) and registered (ACTRN12622000841774). The double blind allowed neither the examiner nor the data analyst to know the allocation to each of the situations.

2.2 Sample size calculation

For the sample size calculation, an alpha error of 0.05 and a beta error of 0.2 and medium effect size (f = 0.25 or Eta partial squared = 0.06) were determined. Therefore, using the G*Power (3.1.9.2) software, at least of 34 participants were required to achieve our objective. To account for an estimated dropout rate of 10%, the required sample size was adjusted to 38 participants, divided into two groups (n = 19).

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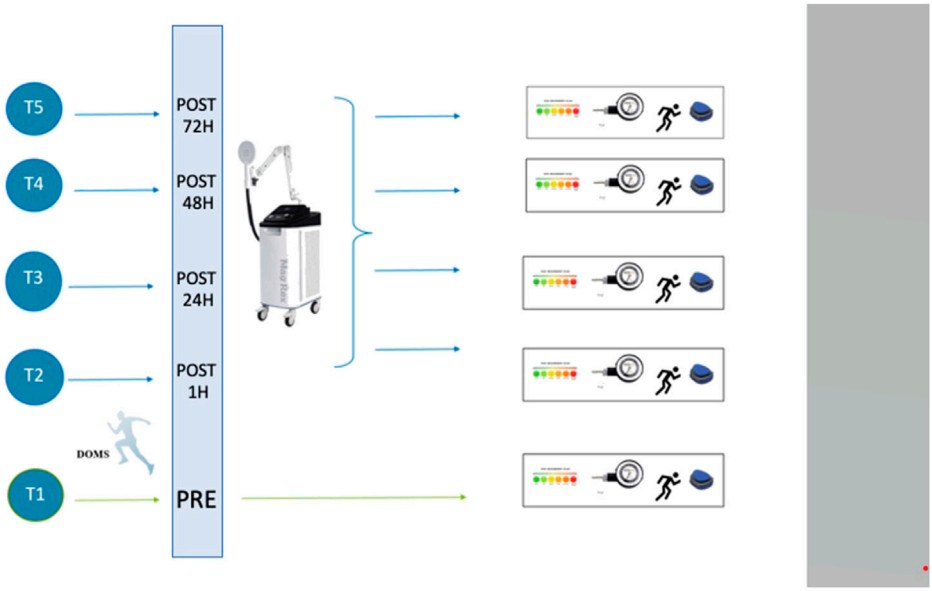

FIGURE 1
Study design and intervention schedule. Pre-Post: times before, after 1 h, 24 h, 48 h, and 72 h of eccentric exercise-induced delayed-onset muscle soreness.

2.3 Participants

In order to be part of the study, the participants had to meet the inclusion criteria: 1) male subjects aged 18 and 35 years old; 2) not have suffered a lower limb muscle injury in the past 6 months; 3) not have undergone surgery in the past 12 months of the lower limbs; 4) not presenting any contraindication regarding the application of electrotherapy. Once participants were enrolled in the study, demographic data (age) and anthropometric (weight and height) data were recorded. The weight (kg) was measured with a scale (Ano Sayol SL, Barcelona, Spain) and the height (cm) with a stadiometer (Asimed T2, Barcelona, Spain). The body mass index (BMI) was calculated as weight (kg)/height2 (m). Duration and number of training sessions was recorded.

The participants were randomized to the intervention group or the placebo group using the random function of Microsoft Office Excel (Microsoft Corporation, Redmond, Washington, United States). Prior to be included in the study, all participants signed informed consent.

2.4 Procedure

The participants were submitted to five evaluation sessions throughout the study, as illustrated in Figure 1.

All participants have a familiarization session 1 week before the first session of the study. On the first session (day one, pre muscle damage protocol) anthropometric, creatine kinase, blood lactate levels, half squat speed, maximal voluntary contraction of the quadriceps, fatigue perception and PPT measurements were performed. On the first session (1 h after), 24, 48 and 72 h after the muscle damage protocol the same measurements were recorded.

In order to examine the effects of the muscle damge protocol the creatine kinase and lactate levels were recorded. Creatin kinase (CK) measurement was performed using venous blood obtained by venipuncture. A diagnostic instrument designed to analyze blood parameters using test strips called Reflotron Plus (Roche Diagnostics, Barcelona, Spain) was used following the study by Baird et al. (2012). Participants were reminded not to engage in physical activity for at least 2 days prior to the study, as this enzyme may rise days after physical exercise. Lactate levels were obtained from venous blood following the protocol of Clemente et al. (2010). Blood lactate concentrations were measured by electroenzymatic analysis (Lactate Scout Pro, Musimedic S.L. Donostia, Spain).

2.5 Intervention

2.5.1 Eccentric exercise protocol

All the participants performed the eccentric exercises session for 60 min. The protocol was based on the study performed by Howatson et al. (2012). They previously performed a warm-up with the physical trainer based on ballistic, mobility, strength, and running exercises. The 60-min session included performing 10 sets of 10 repetitions of vertical countermovement high jump from a 40-cm-high step (drop jump), carrying 10% bodyweight dumbbells, with a 1-min rest between sets, and 15 s between each repetition. In addition, a circuit of eccentric exercises was added, which is detailed below. After the plyometric exercise, 3 sets of the 3 exercises described below were performed: 1) Lower limbs rear kick (one-leg deadlift) hanging 10 kg of weight on the chest, performing 3 sets of 10 repetitions with each leg. 2) Suspension race simulation, performing 3 sets of 20 s 3) Nordic hamstring curl, performing 3 sets of 10 repetitions.

2.5.2 Peripheral electromagnetic stimulation

After the induced DOMS session, the participants received a peripheral electromagnetic stimulation treatment with a Long-Term Potentiation (LTP) program described below, in the case of being part of the experimental group (EG). As for the control group (CG), the machine was placed 20 cm far from the skin to make sure that the effect could not reach it, and with a lower intensity so that the participants heard the same noise as that of their teammates in the EG, but without finding an effect.

This treatment was based on the studies of Sdrulla et al. about the improvement of synapses with the use of pulse trains of 100 HZ on the Aβ fibers. This stimulation also produces an effect at the central level (Beltrá et al., 2022). It consisted of 5 stimulations of 5 s at 100 HZ with 55 s of rest for a total of 5 min of treatment (Sdrulla et al., 2015). It was applied every day during the 3 days just before the measurements. The probe was placed on the inner side of the thigh near the femoral nerve to act on the entire quadriceps.

2.6 Outcome measures

2.6.1 Primary outcome measure

2.6.1.1 Pressure pain threshold

The protocol and methodology of the study was conducted following previously performed by Fleckenstein et al. (2017). It is currently known that the pressure pain threshold (PPT) is a valid and reliable method to assess mechanical threshold. It can be defined as the minimum pressure necessary to produce pain. Before starting the eccentric exercises, measurements had to be made to verify that the participants did not experience pain on palpation and the concept of PPT was explained to them, as "the moment in which the pressure stimulus changes to a sensation of pressure at pain". The applied pressure was 1kg/cm2, until the subjects reported the onset of pain. In this study a Baseline algometer (Wagner instruments, Greenwich, United States) was used.

2.6.2 Secondary outcome measures

2.6.2.1 Visual Analogue Scale—Fatigue

The Visual Analog Scale to evaluate fatigue (VAS-F) was used to evaluate fatigue during the HS test (Nadarajah et al., 2017). Before carrying out the performance analysis test, the participants were informed that they would be asked at the end of the jump test the amount of pain that they could feel during the execution of the half squat (HS) test. When doing so, it is necessary to show them a 100 mm horizontal line, of this size to better appreciate the changes throughout the study, that the extremes represent the extremities of the evaluated fatigue: "0"will be the absence of fatigue and "100" an unbearable fatigue. The participants had to indicate their perception of fatigue after performing the HS test. This instrument has showed an excellent internal consistency (Cronbach alpha = 0.90) (Nadarajah et al., 2017).

2.6.2.2 Half Squat

Lower body performance was analyzed with a HS test consisting of maximum power effort. A warm-up was previously carried out for the familiarization of the exercise, for the angulation of the hip and knee joint. The HS was previously performed at 50% of the participant's body weight, then the missing kilograms were added to reach 80% of the body weight of each subject. Next, the maximum mechanical power developed with an added load (80% of body weight) was evaluated with the HS exercise. They had to perform 2 repetitions with 3 min of rest between them. To fix the load that corresponds to 80% of each subject, it was necessary to carry out a test on the first day in the gym. This load was then maintained throughout the study. When perfoming the HS test, participants stand in a standing position with their feet hip-width and shoulder-width apart. The hands must be gripped to the bar, which was supported later by the cervical spine, at the C5-C7 level on the trapezius and rear deltoids, with a protective ergonomic cushion. Participants must flex their knees to 90° and then proceed to full power extension to return to the original standing position (Bogdanis et al., 2014). The test was carried out at maximum speed on a Smith Multipower bar guide machine, using 20, 10, 5, 2.5 and 1.25 kg discs (Technogym, Gambettola, Italy). By having both ends fixed, this allows only the movement of the bar, vertical in this case. To have an estimation of the execution speed of each subject in each repetition, the test was carried out with an Encoder of the ADR encoder brand (Toledo, Spain). It is held attached to one end of the bar so as not to disturb the movement of the HS. Thus, it allows to measure the vertical displacement of the bar in terms of the movement of the exercise, implementing the software of the ADR system (Toledo, Spain), and recording the speed of the bar in m/s (average and peak).

2.6.2.3 Quadriceps strength

A hand-held dynamometer (Activbody, San Diego, CA) (Thorborg et al., 2009) was used to assess maximum voluntary isometric contractions of quadriceps (Hung et al., 2021). A total of three repetitions with the maximum power of each participant was measured (Hung et al., 2021). The participants were seated on the edge of a physiotherapy table with the knee at 90 degrees of flexion, and the dynamometer was placed at the level of distal tibia (Hung et al., 2021). Then, the maximum peak force during each contraction was sought with a normalization against the weight of each participant (Rezaei et al., 2014). There was a rest of 3 minutes between measures (Babault et al., 2003; Rezaei et al., 2014). At the end of the measurement, the participants rested for 3 min before continuing with the other measurements.

2.7 Statistical analysis

Statistical analyses were carried out using the SPSS v.25 program (IBM, Armonk, NY, United States). The normal distribution of the data was verified using the Shapiro-Wilk test and histograms. p values < 0.05 were considered as non-normally distributed and $p > 0.05$ as normally distributed a descriptive analysis was conducted to characterize the sample. Central tendency and dispersion data were reported as mean and standard deviation for normal-distributed variables, or as median and interquartile range for non-normal-distributed variables, respectively. For the quantitative variables, independent t-test or Mann-Whitney U test were carried out to compare the means groups at baseline. Homoskedasticity and sphericity were checked. When the assumptions were met, a mixed analysis of variance (ANOVA) 5×2 was carried out, adjusting multiple comparisons with the Bonferroni test. The effect size was estimated with partial eta squared (η 2), interpreting values of .01, .06, .14 as small, medium and large, respectively. A 95% confidence interval was followed (Adams et al., 2021).

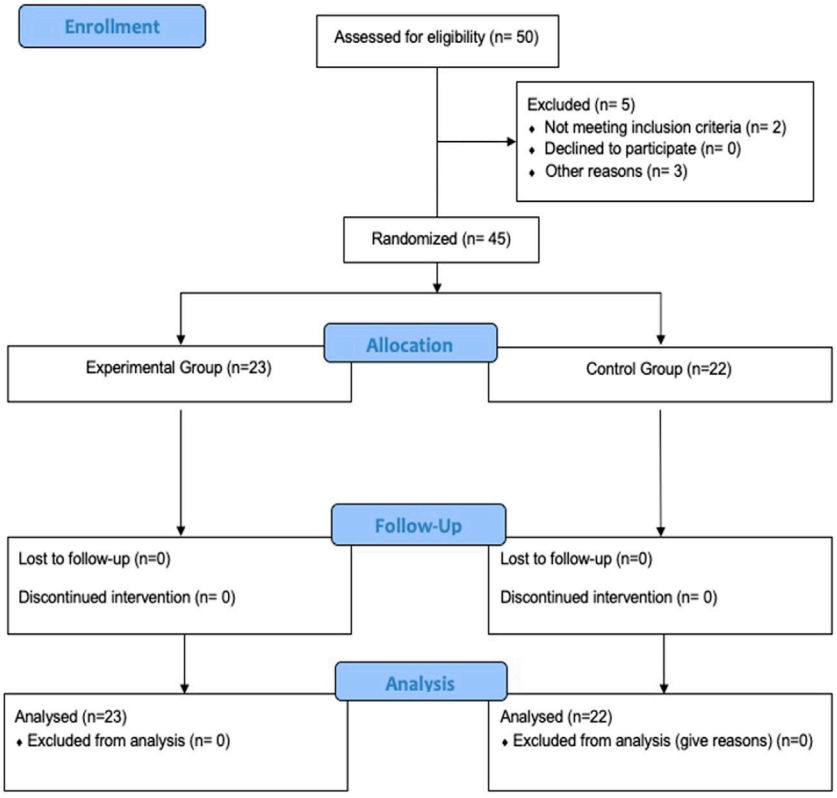

FIGURE 2
Study flowchart.

TABLE 1 Sociodemographic characteristics and pre-treatment scores of the total sample, experimental and control group.

Variables	Total sample (n = 45)	EG (n = 23)	CG (n = 22)	*p*-value (between-group)
Age (years)	22.33 ± 4.42	22.87 ± 4.94	21.77 ± 3.84	0.412
Height (cm)	178.67 ± 6.88	178.39 ± 7.27	178.96 ± 6.62	0.787
Weight (kg)	74.14 ± 8.83	74.76 ± 8.29	73.50 ± 9.52	0.636
Body mass index (kg./m2)	23.21 ± 2.24	23.48 ± 2.13	22.92 ± 2.36	0.405
Soccer practice experience (years)	14.52 ± 5.6	14.54 ± 5.56	16.12 ± 3.83	0.890

CG, control group; EG, experimental group. Results are expressed as mean ± standard deviation

3 Results

3.1 Baseline data

Out of a total of 50 volunteers willing to participate in the study, 5 were excluded (Figure 2). Therefore, a total of 45 male participants aged 22.33 ± 4.82 years were included and analyzed in the study. The anthropometric characteristic did not differ between participants of both groups (Table 1). All the participants trained five times a week for 90 min.

To control that the response to the eccentric exercise-induced DOMS session was homogeneous for both groups, the CK and blood levels of lactate were measured. An increased of CK was observed 1, 24, 48 and 72 h after intervention in both groups ($p < 0.001$) and no interaction between groups was detected in pre ($p = 0.589$), post 1 h ($p = 137$), post 24 h ($p = 0.891$), post 48 h ($p = 0.284$) and post 72 h ($p = 0.379$). An increased of blood lactate levels was observed 1 h after intervention in both groups ($p < 0.001$) and no interaction between groups and time was observed ($p = 0.814$) (Figure 3).

Baseline measures of the dependent variables did not show significant differences between experimental and control group (Table 2). There was no group-by-time interaction for HS, ($p = 0.380$), quadriceps strength ($p = 0.652$) and VAS-F ($p = 0.864$). Group-by-time interaction was observed in PPT of the vastus medialis ($p = 0.040$) with a medium effect size ($\eta^2_p = 0.069$) (Table 2). From 48 to 72 h the experimental group showed an increase of PPT compared to control group ($p = 0.015$) (Figure 4). Time effect was observed for HS, quadriceps strength, VAS-F and

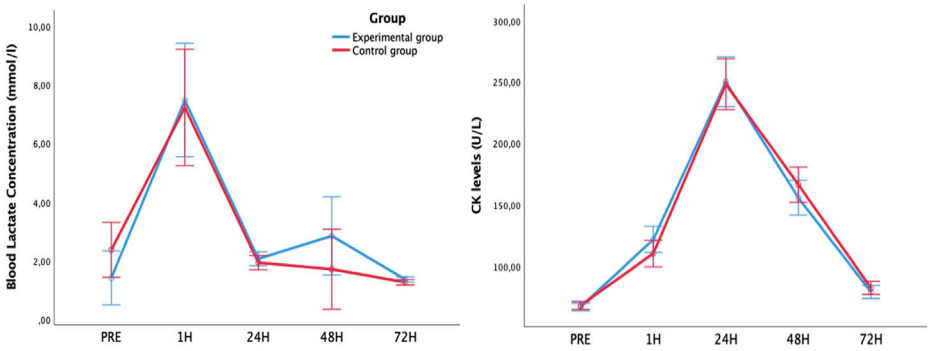

FIGURE 3
Blood Lactate (left) and creatin kinase (right) concentration levels evolution after eccentric exercise-induced delayed-onset muscle soreness comparing experimental and control group.

TABLE 2 Differences between the experimental and control groups for the dependent variables.

	T1 (baseline)	T2 (post 1 h)	T3 (post 24 h)	T4 (post 48 h)	T5 (post 72 h)	p (time x group)	η^2_p
Primary Outcome Measure							
PPT (kg/cm2)							
CG (n = 22)	10.77 ± 1.47	4.89 ± 2.61	6.82 ± 4.62	5.99 ± 5.22	7.14 ± 4.58	0.040	0.069
EG (n = 23)	10.83 ± 1.65	5.96 ± 2.19	6.37 ± 3.09	7.17 ± 3.83	9.74 ± 5.76		
Secondary Ourcome Measures							
Quadriceps Strength (Kg)							
CG (n = 22)	43.99 ± 14.23	36.84 ± 12.45	42.75 ± 12.15	47.22 ± 11.21	49.60 ± 11.76	0.652	0.010
EG (n = 23)	47.96 ± 11.73	39.04 ± 10.91	44.59 ± 11.19	47.96 ± 12.58	50.98 ± 13.07		
VAS-F (mm)							
CG (n = 22)	0.00 ± 0.00	49.59 ± 16.95	44.32 ± 20.84	34.68 ± 22.87	21.23 ± 15.00	0.864	0.007
EG (n = 23)	0.00 ± 0.00	50.91 ± 16.67	43.48 ± 14.65	37.30 ± 19.83	26.09 ± 18.09		
Half Squat (m/s)							
CG (n = 22)	0.81 ± 0.11	0.68 ± 0.12	0.73 ± 0.11	0.77 ± 0.12	0.80 ± 0.11	0.365	0.024
EG (n = 23)	0.85 ± 0.12	0.74 ± 0.08	0.75 ± 0.10	0.81 ± 0.14	0.86 ± 0.11		

CG, control group; EG, experimental group; PPT, pain pressure threshold; VAS, visual analogic scale to evaluate fatigue; η^2_p, partial eta squared.

PPT ($p < 0.001$). Despite a progressive recovery of maximal strength of the quadriceps after the muscle damage protocol, HS speed, PPT of the vastus medialis did not return to baseline values within 72 h (Table 2; Figure 4).

4 Discussion

This novel study aimed to observe the influence of peripheral electromagnetic stimulation on performance and recovery in professional soccer players suffering from induced-DOMS. Peripheral electromagnetic stimulation aimed to improve the PPT in the quadriceps, compared to control group. There were no significant differences between groups and time in the rest of the variables analyzed.

Including eccentric exercise sessions that cause DOMS in training dynamics is necessary in sports to improve performance (Cheung et al., 2003). Several articles have been published that explore the strategies offered by electrotherapy by applying a TENS for recovery that aims to accompany the athlete and minimize indirect markers of muscle damage (Craig et al., 1996; Butera et al., 2018). Our study evaluated the effects of applying the LTP protocol in males who suffered from DOMS induced by a 1-h session of eccentric strength exercises, never tested to date.

According to the results, the group treated with the peripheral electromagnetic stimulation current did not improve the muscle recovery in terms of performance in the 72 h after performing a series of exercises with large eccentric component that generated DOMS, compared to the control

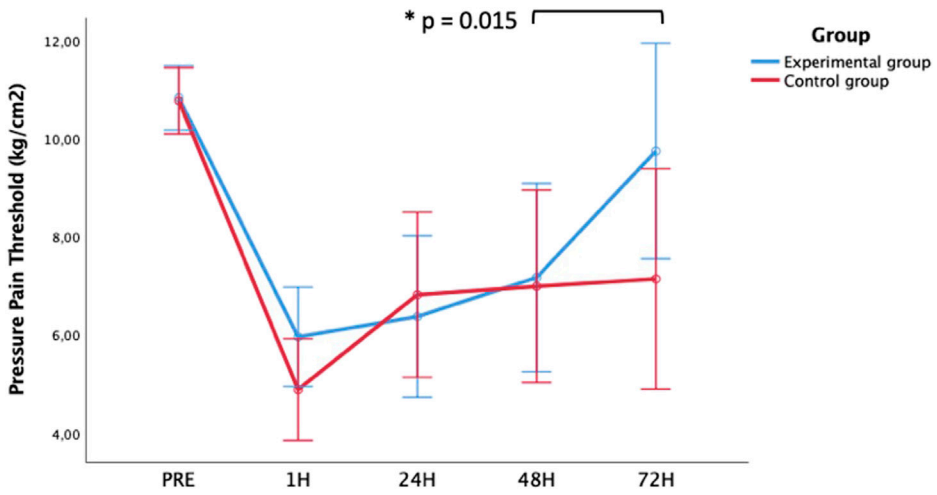

FIGURE 4
Blood Lactate (left) and creatin kinase (right) concentration levels evolution after eccentric exercise-induced delayed-onset muscle soreness comparing experimental and control group.

group. These findings would indicate that the muscular responses that follow this type of acute exercises that cause DOMS were not altered by the treatment, since in both groups showed a decrease in performance in the measurements made in the post exercise/T2. The improvements observed in the PPT of the vastus medialis could have been influenced by other mechanisms, such as peripheral sensitization (Amiri et al., 2021; García-Pérez-de-Sevilla et al., 2022).

First, pressure pain threshold was analyzed with an algometer at the vastus medialis of the quadriceps, revealing a significant difference between groups favoring the EG, with a moderate effect size, even considering that there is a physiological process where after 48 h the PPT is usually higher due to the regeneration processes that are taking place in the damaged fibers (Fleckenstein et al., 2017). In addition, there was a significant difference between groups in the PPT of the vastus medialis comparing T4 and T5 favoring the peripheral electromagnetic stimulation (p = 0.015), as has been observed in other studies (Sdrulla et al., 2015). Peripheral electromagnetic stimulation also has anti-inflammatory effects. Inflammation is a natural response of the body to injury or infection, but when it becomes chronic, it can cause tissue damage and contribute to the development of musculoskeletal conditions. Peripheral electromagnetic stimulation has been shown to reduce inflammation in several ways. First, it increases blood flow to the affected area, which helps to remove inflammatory chemicals from the tissue. Second, it stimulates the production of anti-inflammatory cytokines, which help to reduce inflammation. Finally, peripheral electromagnetic stimulation has been shown to decrease the production of pro-inflammatory cytokines, which contribute to the development of chronic inflammation (Barker, 1991).

In the analysis of the perception of fatigue measured with the VAS-F scale, no significant differences were found between groups. Fan and Sdrulla showed that distal hypoalgesia is not seen in all the stimuli that athletes may face when training (Fan and Sdrulla, 2020). In addition, other muscle groups apart from the quadriceps could have had an influence on the pain perception described by the participants.

Regarding the HS test, there were no significant differences between groups, which may be due to the synergy of other muscles that are involved in performing this test. Similar findings were reported in the study performed by Goulart et al. in which they suggested that the interval between sessions could influence the performance of explosive actions, although the HS test is a validated tool to estimate the power of the lower limbs (Goulart KN de et al., 2021). On the other hand, in the study conducted by Dominguez-Balmaseda et al. the EG maintained the power of the lower limbs in HS after performing a protocol of eccentric exercises that induce DOMS, especially in the post 24 h. It should be noted that the study carried out by Dominguez-Balmaseda et al. had a larger sample size, and the treatment consisted of herbal supplementation (Dominguez-Balmaseda et al., 2020). Although in the present study there were no significant between-group differences in the power of the lower limbs, there were significant improvements in the EG in the within-group analysis, suggesting that it would be interesting to perform future studies with a larger sample size to see if this therapeutic strategy could improve DOMS symptoms and with it, sports performance.

Concerning the quadriceps strength, there were not significant differences between groups, but this could be influenced by the loss of strength that the test entails if the participant approaches the extension (Babault et al., 2003). However, there were significant differences in the within-group analysis for the EG, suggesting that the peripheral electromagnetic stimulation current should be further investigated. Other post-exercise recovery strategies such as electrostimulation or protein ingestion has been proposed. Recently, Hartono et al. (2022) have performed a systematic review including 84 trained young males and ten longer-term studies including 167 trained and 391 untrained participants. They conclude that protein ingestion enhanced myofibrillar protein synthesis rates, but not mitochondrial protein synthesis rates during post-exercise recovery after an acute bout of

concurrent exercise demonstrated in acute studies. Of the included longer-term training studies, five out of nine reported that protein supplementation enhanced concurrent training-mediated increases in muscle mass, while five out of nine studies reported that protein supplementation enhanced concurrent training-mediated increases in muscle strength and/or power. Therefore, the peripheral electromagnetic stimulation may not increase the muscle strength or peak force after an eccentric exercise-induced delayed-onset muscle soreness protocol compared to the control group. Protein ingestion or other potential recovery strategies could be enhanced to improve the performance in soccer players.

4.1 Limitations

The present study has several limitations. A key factor that we should consider is that the usual load of the training sessions followed by the participants could influence their recoveries as they are used to higher intensity work compared to the exercises performed in our study. Finally, the findings of the present study cannot be extrapolated to female or sedentary population. Therefore, it would be interesting to evaluate the efficacy of our treatment in other types of populations like female participants or less active populations.

In future studies, parameters such as heart rate variability or the Counter Movement Jump test should be evaluated, due to the already demonstrated implication that both have on recovery and anticipation by athletes. In addition, parameters such as electromyography (EMG) could be added in future research to be able to analyze muscle recruitment after peripheral stimulation, to determine the real impact of peripheral stimulation on the peripheral nervous system (Beltrá et al., 2022). It would be interesting in future studies to interact with athletes of the same competitive level at the beginning of the season in order to see if the daily training load influences recovery. Clemente et al. carried out a review in 2021 where they observed that soccer players present lower physical performance at the beginning of a pre-season. Thus, it would be interesting if future studies are developed in the early stages of a return to training (Clemente et al., 2021). In addition, it would be interesting to add the 30-m run, since it is a common sprint distance in soccer, to deepen the analysis of the influence of peripheral stimulation on physical performance (Pearcey et al., 2015).

4.2 Practical application

Peripheral electromagnetic stimulation is a safe and effective therapy for musculoskeletal conditions. It can relieve pain, reduce inflammation, improve circulation, and promote tissue healing. Peripheral electromagnetic stimulation can be used alone or in combination with other therapies, such as physical therapy, cryotherapy or supplementation.

5 Conclusion

Peripheral electromagnetic stimulation in male professional soccer players did not produce significant improvements in the power and strength of the lower limbs but decreased the peripheral sensitization measured by pressure pain threshold of the vastus medialis after an eccentric exercise-induced delayed-onset muscle soreness protocol.

Data availability statement

The raw data supporting the conclusion of this article will be made available by the authors, without undue reservation.

Ethics statement

The studies involving human participants were reviewed and approved by the Comité de INvestigación. Universidad Europea de Madrid. The patients/participants provided their written informed consent to participate in this study.

Author contributions

AS and HK carried out the design and idea of the project, AG-F, HK, and DB wrote the introduction to the manuscript, AG-F, GP, and DM wrote the methodology and statistics part, AG-F, DB, and HK wrote the Discussion and conclusions part. All authors contributed to the article and approved the submitted version.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Crossover gait in running and measuring foot inversion angle at initial foot strike: a front-view video analysis approach

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Introduction: Foot inversion angle at initial foot strike is associated with various running-related injuries. Traditionally, video analysis of foot inversion angle has been accomplished by positioning a camera to record from the back view, but complications arise when a crossover gait obscures the area of measurement. This study aims to investigate the viability of measuring foot inversion angles at initial foot strike of running from the front view as an alternative to using the back view in 2D video analysis.

Methods: Forty-four healthy runners (20 females, 24 males) ran at their self-selected speeds on a treadmill with their gait recorded from front and back camera views. Foot inversion angles at initial foot strike were analyzed using Kinovea. A 2 × 2 (Camera × Foot) ANOVA with repeated measures was performed on the foot inversion angle data. Subsequently, correlation and linear regression were performed to determine the relationship between the back and front-view measurements.

Results: Thirteen runners (29.5%) displayed crossover gait within 18 gait cycles. ANOVA revealed a significant main effect on Camera ($p < .001$) only, where foot inversion angle was greater from the front camera view. Correlation analysis showed a significant positive correlation between the front and back camera views ($r = 0.388$, $p < .001$). Regression analyses yielded an equation, $y = 0.42 + 0.53x$, where y and x were the foot inversion angle measured from the back and front camera views, respectively.

Discussion: With a linear regression conversion equation, front-view foot inversion angles at initial foot strike can be used to determine rearfoot inversion angles when crossover gait obstructs the back camera view.

KEYWORDS

running-related injuries, 2D motion analysis, low-cost alternative, foot kinematics, gait analysis

1 Introduction

Running is a common form of physical activity that provides many health benefits and has attracted more and more participation in recent years (Vitti et al., 2020). However, running has been considered a high-risk physical activity due to the high incidence rate of overuse running-related injuries (RRIs) reported among novices (Kluitenberg et al., 2015; Kemler et al., 2018) and recreational runners (Hespanhol Junior et al., 2013). Novice runners are classified as individuals who have less than 1 year of regular running experience while recreational runners are classified

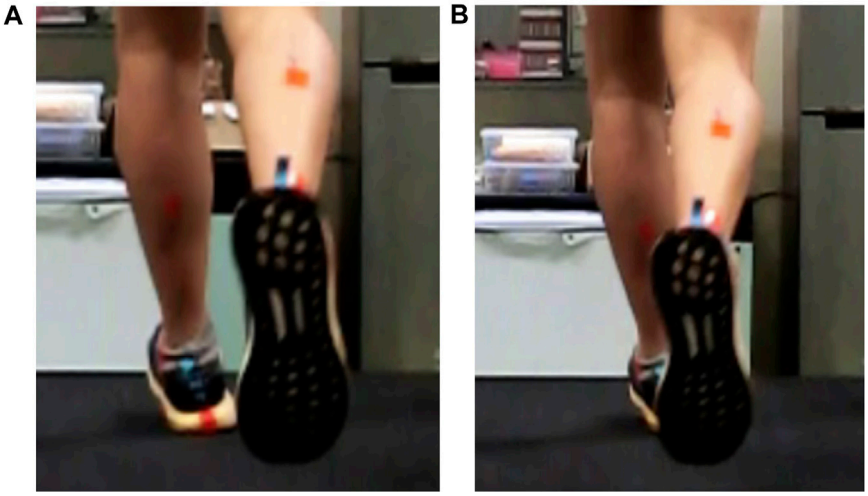

FIGURE 1
Example of foot inversion at initial foot strike captured from a back camera view. The initial foot strike from the back view is **(A)** unobstructed, and **(B)** blocked by the right foot.

as individuals who have been engaging in running activities minimally once per week for 1 year (Honert et al., 2020). Notably, a systematic review found that novice runners face a considerably greater risk of injury, with a rate of 17.8 injuries per 1,000 h of running, whereas recreational runners experienced 7.7 injuries per 1,000 h of running (Videbæk et al., 2015). In view of this, studies have sought to establish the biomechanical factors of various RRIs via gait analysis (Fatone & Stine, 2015; Ceyssens et al., 2019), to gain insights into the running techniques that may increase the injury risk. Thereafter, an appropriate rehabilitation or gait retraining program can be advised (Roper et al., 2016; Warne et al., 2017).

Gait analyses are commonly performed using either a two-dimensional (2D) video-based or three-dimensional (3D) motion capture system. Although 3D systems have been considered the gold standard, it requires expensive specialized equipment and would require some level of expertise to operate. Additionally, data collections are tedious and limited to laboratory settings. With these constraints, clinicians and researchers have explored the use of 2D video analysis (Damsted et al., 2015; Fatone et al., 2015; Hensley et al., 2021), which is relatively simple and of low cost (Fatone and Stine, 2015). Recorded videos can subsequently be integrated with commercially available video analysis software (e.g., Dartfish, Kinovea, and Hudl) that allow users to annotate at specific key points of interest (DeFroda et al., 2016) to obtain kinematic measurements. The portability of 2D video analysis also makes it a cost-effective tool for practitioners. In a recent study, Hensley et al. (2021) found that about 75% of practitioners use 2D video cameras during their clinical gait assessment. This highlights the ease of accessibility and popularity to conduct gait analysis using 2D video analysis among practitioners.

Previous studies have demonstrated moderate to high correlations between 2D video analysis and 3D motion capture when evaluating gait parameters in running (Maykut et al., 2015; Dingenen et al., 2018b; De Oliveira et al., 2019; Martinez et al., 2022). Specifically, Martinez et al. (2022) observed sagittal plane kinematic differences of 2.5°–5°, while Maykut et al. (2015) reported differences of 1°–4° for frontal plane kinematics. Multiple studies have also consistently reported excellent intra and inter-rater reliability in 2D analysis with an intra-class correlation coefficient exceeding 0.90, indicating strong reliability in 2D analysis (Maykut et al., 2015; Dingenen et al., 2018b; De Oliveria et al., 2019; Mousavi et al., 2020). Thus, these studies have shown compelling evidence that the application of gait analysis using 2D video analysis is possible.

Despite the aforementioned advantages of 2D video analysis, there are certain limitations when using it for gait analysis. One such limitation is ensuring the visibility of key landmarks at all times for post-processing analysis. In situations where occlusions of key landmarks occur, it may hinder clinicians and researchers to obtain measurements at a specific gait cycle event. In particular, the rearfoot inversion angle at the initial foot strike, typically measured from a back camera view (Cornwall and McPoil, 1995), is known to obstruct the visibility of key landmarks during post-processing analysis. The initial foot strike is the instance when the foot first makes contact with the ground which also marks the beginning and end of a gait cycle (Novacheck, 1998). A problem arises when runners display a crossover gait, where the planted foot is obstructed by the other foot in the swing phase (Figure 1). This phenomenon was also observed by Vincent et al. (2014). The rearfoot inversion angle at the initial foot strike holds significant importance as it is associated with RRIs such as iliotibial band syndrome (Grau et al., 2008) and Achilles tendinopathy (McCrory et al., 1999; Mousavi et al., 2019). Furthermore, the measurement of this angle is essential for evaluating other variables related to RRIs such as ankle eversion range of motion, eversion velocity, and eversion duration during the stance phase of running (Kuhman et al., 2016). However, the presence of a crossover gait in runners may hinder clinicians and researchers from obtaining accurate measurements of the foot inversion angle at the initial foot strike when recorded from the back camera view. Hence, an alternative camera view may help evaluate such gait variables.

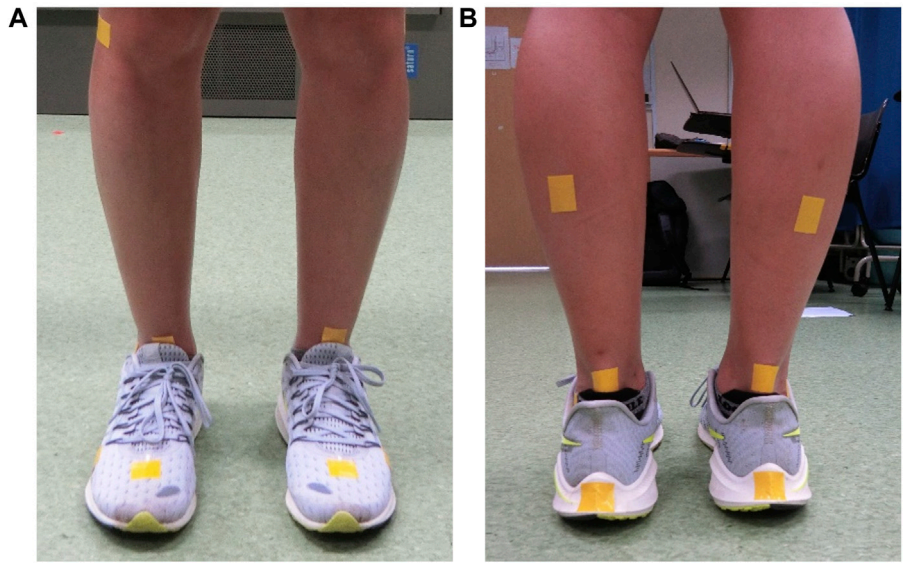

FIGURE 2
Marker placement using cloth tapes in **(A)** the front view and **(B)** the back view. Other markers not mentioned earlier are not relevant to this study.

Front-view video analysis poses a possible method to overcome the crossover gait problem since the obstruction only occurs in the dorsal plane. However, it remains unclear whether the measurements obtained from front and back views can be compared. The current study aimed to investigate the efficacy of using a front camera view in analyzing foot inversion angle during treadmill running. Based on a previous study that reported a positive correlation between static forefoot varus and rearfoot angle (Johanson et al., 2010), it was hypothesized that the foot inversion angle measured from front and back camera views would be comparable. The findings would be expected to provide valuable insights to clinicians and researchers in evaluating foot inversion angle among runners with crossover gait.

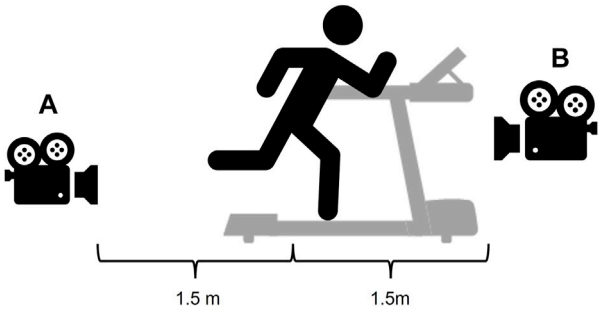

FIGURE 3
Illustration of video camera placements for recording in the **(A)** back camera view and **(B)** front camera view.

2 Materials and methods

2.1 Participants

Forty-four (20 females, 24 males) healthy recreational runners (age 25.2 ± 6.5 years; height 167.2 ± 7.0 cm; body mass 62.8 ± 8.1 kg) were recruited. The inclusion criteria for the study were: (1) individuals who were recreational runners, with more than 1 year of regular running months (Honert et al., 2020), (2) with a minimum of one weekly run for the past three, (3) between the ages of 18–45, (4) comfortable to run on a treadmill without the need for handrails, and (5) capable of running continuously at a speed of 8 km/h for at least 10 min (Gazendam & Hof, 2007). Participants were excluded from the study if they (1) were pregnant during the time of the study (for female participants), (2) answered positively to any questions on the Physical Activity Readiness Questionnaire (PARQ+ 2020) indicating the presence of serious health conditions, (3) had undergone surgery on the legs within the past year, or (4) had lower limbs injuries requiring more than 7 days of rest in the past 6 months. All participants provided written informed consent prior to their participation and ethics approval was granted by the Nanyang Technological University Institutional Review Board (IRB-2021-124).

2.2 Experimental setup

All participants were invited to run at their comfortable paces on a treadmill (h/p/cosmos saturn®, h/p/cosmos® sports & medical gmbh, Nusseldorf-Traunstein, Germany) using their own running shoes. To facilitate video analysis, cloth tapes were placed on the anterior and posterior talus, calcaneus, and mid-gastrocnemius on both limbs in each participant (Figure 2). The study protocol commenced with a 5-min warm-up session (Riley et al., 2018) where participants were allowed to transition from walking to running. Immediately after, the participants were directed to incrementally increase their

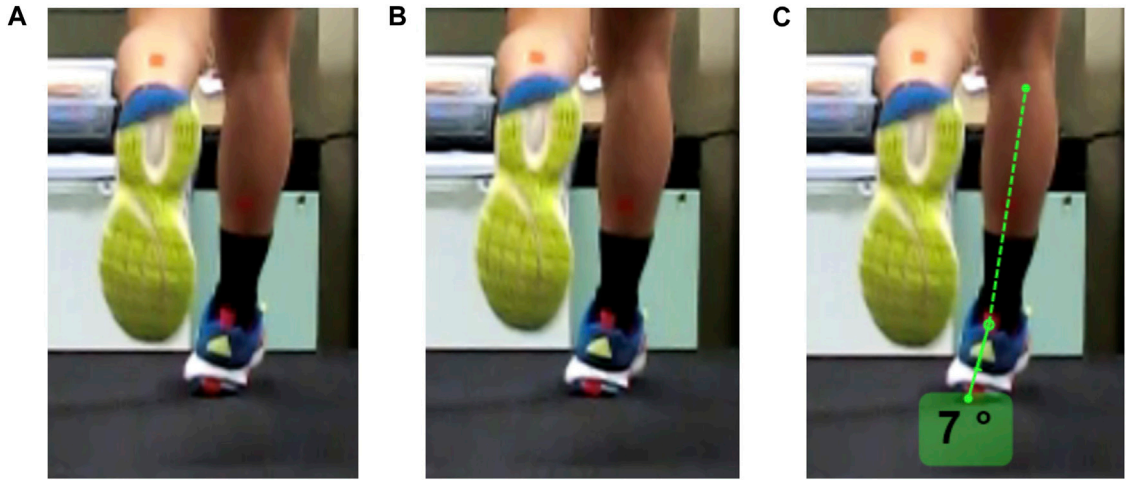

FIGURE 4
Frame-by-frame shots from the back view. **(A)** 1 frame before the initial foot strike, **(B)** frame of initial foot strike as heel contact is just made with the treadmill, and **(C)** angle measurements are derived from the line of posterior talus to the middle of gastrocnemius and calcaneus. Positive values indicate inversion while negative values indicate eversion.

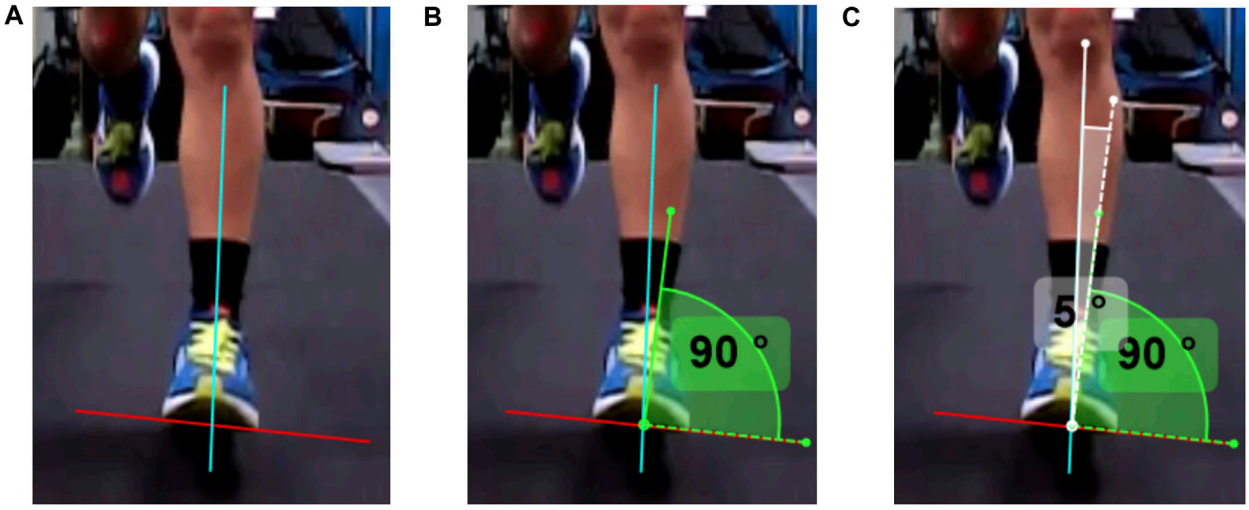

FIGURE 5
The sequence of steps taken to measure foot inversion at the initial foot strike from the front camera view. **(A)** In step 1, a parallel line (red) to the forefoot is drawn and a line (blue) from the base of the foot to the anterior talus. **(B)** In step 2, a right-angled (90°) measurement is placed at the intersecting point towards the lateral aspect of the foot. **(C)** In step 3, the angle is measured between the blue line to the relative right-angled measurement. Positive values indicate inversion while negative values indicate eversion.

running speed to a self-selected comfortable pace (9.4 ± 1.1 km/h) which they had to maintain for approximately 10 min. Verbal confirmation of their comfortable running speed was obtained prior to the start of the video recording. Participants were also instructed to maintain their running position at a consistent area of the treadmill. For recording the participants' running, a single video camera (Casio Exilim EX-100) was utilized, and the camera was set at different spots for video recording from the front and back view. The video recordings were captured sequentially, with the front view being recorded first, followed by the back view.

The video camera was placed 1.5 m away, perpendicular to the participants' coronal plane (Figure 3). The camera height was set at the hip (front view) and ankle (back view) levels respectively for each participant. Each camera view was recorded for approximately 30 s at 120 Hz while participants were running at their self-selected speed during the 10-min window. Participants continued running while researchers move the camera position. Multiple recordings were taken in each camera position, and the video in which participants displayed minimal movement (moving sideways, etc.) was selected by the researchers for analysis.

Souza (2016) provided guidelines that were followed by researchers to standardize the identification of the initial foot strike frame. After identifying the frame of initial foot strike, the researchers used the following procedures to derive angle measurements for each camera view. In the back camera view, the inversion angle was calculated between the line extending from the posterior talus to the midpoint of the gastrocnemius and calcaneus (Figure 4). In the front camera view, the inversion angle was calculated by measuring the angle formed between the line from the base of the foot to the anterior talus and the perpendicular angle from the base of the foot (Figure 5). Positive values indicate inversion, while negative values indicated eversion in both measurement methods. Figures 4, 5 outline the steps taken in detail for angle measurements in the front and back camera views, respectively. Subsequently, 18 consecutive gait cycles of each foot from each camera view were analyzed in Kinovea (Version 0.8.27). The mean values across the 18 gait cycles were used for analysis. The reason for selecting 18 gait cycles was based on previous research indicating that a minimum number of steps, ranging between 7 and 25 gait cycles, were necessary to attain stable kinematic mean values (Dingenen et al., 2018a; Oliveira & Pirscoveanu, 2021).

2.3 Statistical analysis

To address the aims of the study, two types of analyses were conducted. Preliminary testing was first conducted to screen for normality and homogeneity of variance. As the assumption of normality was violated in the back view data according to the results of Shapiro-Wilk test, all data were log-transformed prior to statistical analysis. A 2 × 2 (Camera × Foot) Analysis of Variance (ANOVA) with repeated measures was performed to compare the measured angles from 2 camera views for each foot. As 2 × 2 ANOVA revealed no significant differences between the left and right feet, data from both feet were combined when performing the Pearson's correlation test. Thereafter, linear regression was performed to derive a single equation for the conversion of the front-view angle to the typical back-view angle. The Wilcoxon signed-rank test was conducted to compare the equation-derived measurements and the actual measurement obtained from the back-view perspective for both the left and right foot. Effect size (r) was interpreted as small ($0.1 \leq r < 0.3$), medium ($0.3 \leq r < 0.5$), or large ($0.5 \leq r \leq 1.0$) (Cohen, 1988). Statistical significance was set at $p < 0.05$ level and the statistical analyses were performed using SPSS (Version 28; SPSS Inc., Chicago, IL, United States) and R (Version 4.1.1). As this study reported discrete values instead of continuous time-series kinematic data, no filtering was performed.

3 Results

The study observed that 13 out of 44 participants (29.5%) displayed at least one crossover gait within the 18 consecutive gait cycles in one foot (Figure 6). Seven participants (15.9%) had crossover gait in both feet while 8 (18.1%) and 12 participants (27.3%) showed crossover gait on the left and right foot, respectively.

Data are expressed as mean (95% confidence intervals). According to the results of the two-way ANOVA test, there was no significant main effect of Foot [$F(1, 43) = 1.773$, $p = .190$, $\eta_p^2 = 0.040$]. The mean angles were similar between the left and right foot in the front [left 10.9° (9.9°, 12.0°), right 11.8° (10.6°, 13.1°)] and back camera views [left 6.1° (4.6°, 7.6°), right 6.8° (5.2°, 8.4°)]. A significant main effect of Camera view was revealed [$F(1, 43) = 62.542$, $p < .001$, $\eta_p^2 = 0.593$], with front camera view measurements [11.4° (10.4°, 12.3°)] being greater than the back camera view measurements [6.5° (5.0°, 7.9°)] by 4.9° [3.5°, 6.3°]. No significant interaction effect was found between the Camera view and Foot [$F(1, 43) = 0.230$, $p = 0.634$, $\eta_p^2 = 0.005$].

Pearson's correlation test revealed a significant positive medium correlation (Figure 7) between the front and back camera view angles ($r = 0.388$, $p < .001$). Linear regression analysis also showed that front camera view measurement accounts for a significant amount of variance for measurements in the back camera view [$F(1, 86) = 15.249$, $p < .001$, $R^2 = 0.151$]. The regression coefficient ($\beta = 0.53$) allowed measurements from the back camera view to be predicted using the regression equation: $y = 0.42 + 0.53\,x$, where y and x were the foot inversion angle measured from the back and front camera views, respectively. The mean angles from the equation-derived measurements are 6.2° (5.7°, 6.7°) for the left and 6.7° (6.1°, 7.3°) for the right foot. Subsequently, Wilcoxon signed-rank test revealed no significant difference between the means of equation-derived measurements and actual measurements for both left ($Z = -0.805$, $p = 0.421$) and right ($Z = -0.140$, $p = 0.889$) feet.

4 Discussion

This study aimed to investigate an alternative approach to measuring foot inversion angle from the frontal view, as opposed to the conventional back camera view measurement (Cornwall and McPoil, 1995). It was initially hypothesized that the front and back view measurements would be comparable. However, the results from this study suggest otherwise as front camera view measurements were significantly greater than back camera view measurements. Therefore, to facilitate comparisons with studies that commonly measure from the back view, an equation was provided to convert the front view measurements to the back view measurement. This conversion enables clinicians and practitioners to effectively utilize the back view measurements for comparative purposes.

4.1 Crossover gait

To provide a gauge on the frequency of crossover gait problems potentially encountered by clinicians and practitioners, the number of crossover gait was reported in this study. Approximately 30% of the participants in the present study displayed a crossover gait, resulting in an obstructed view of their foot when recording from the dorsal plane. This prevents practitioners from accurately measuring the foot inversion angle at the point of the initial foot strike when using the conventional technique. The frequent occurrence of crossover gait reaffirms that the traditional back-view measurement is insufficient and hence there is a need to look for

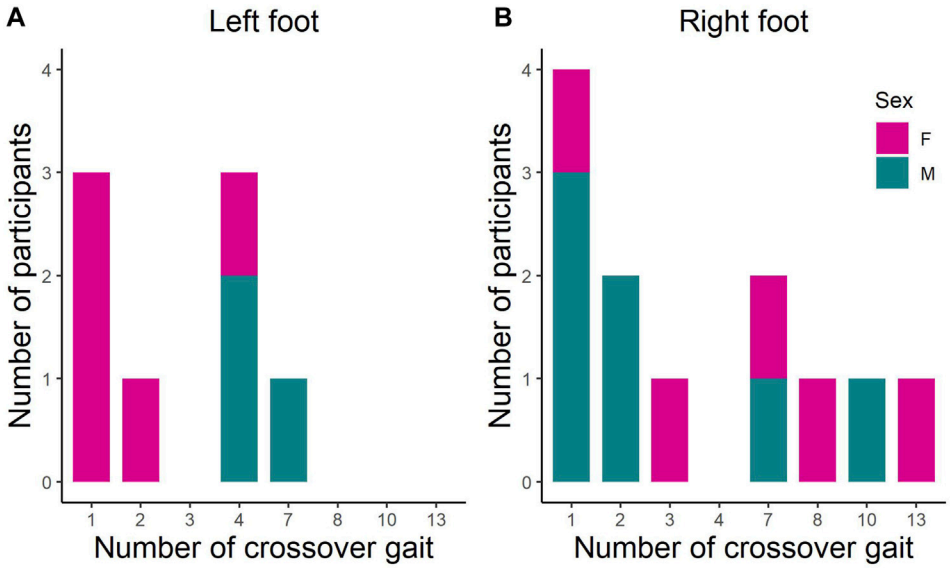

FIGURE 6
The number of crossover gait across 18 gait cycles in the **(A)** left foot and **(B)** right foot in male (M) and female (F) recreational runners.

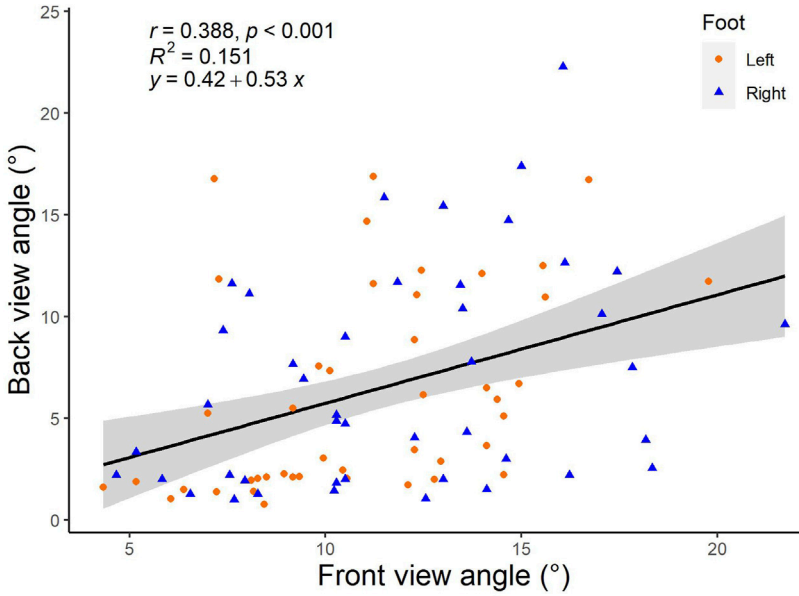

FIGURE 7
Correlation and regression analyses of the front-view and back-view foot inversion angles.

alternative measurements, such as the front-view measurement proposed in our study.

4.2 Foot inversion angle

This study found that the mean front and back camera view measurements were 11.4° (10.4°, 12.3°) and 6.5° (5.0°, 7.9°), respectively. The traditional back camera view measurements were comparable with those obtained by McCrory et al. (1999), who also used a 2D video analysis approach and found that healthy runners displayed rearfoot eversion angles of −7.0° ± 0.7°. These results were similar to Grau et al. (2008), where a 3D motion analysis involving healthy runners revealed foot eversion angles of 1.9° ± 2.6°. In the studies mentioned above, the negative values denote an inversion angle and were measured from the back camera view.

This emphasizes the caution that should be taken in using the values interchangeably. In addition, it also highlights the need for a standard conversion equation to be established, enabling front-view foot measurements to be translated and used in clinical diagnosis.

There are a few possible reasons for the differences in foot inversion angle measurements between the front and back camera views. For simplicity, this study treated the foot as a single rigid body. However, it is understood that the foot is a complex structure in which the forefoot and rearfoot may behave independently (Davis, 2004). This implies that the forefoot may have inverted without affecting the rearfoot. Therefore, the measurements could have differed significantly due to the independent foot structures.

No significant differences were revealed between the left and right foot inversion angles, suggesting that participants utilized comparable foot-landing techniques for both feet. This study did not examine individual foot strike patterns. However, previous research has shown that these techniques can impact rearfoot and forefoot kinematics. For example, heel-strikers are more likely to display greater inversion angles at touchdown compared with forefoot strikers (Pohl & Buckley, 2008). As such, this may have contributed to the discrepancies in measurements between the two camera views.

4.3 Regression analysis

The present study performed a regression analysis to obtain an equation that allows one to easily convert the front-view foot inversion angles to the traditional back-view rearfoot inversion angles. This regression equation can assist practitioners and clinicians in the event of an obstruction of the back camera view. For example, if a runner's foot inversion angle is measured to be 10° from the front view, clinicians can calculate the supposedly back view angle to be approximately 5.7° (0.42 + 0.53 × 10°) and thereafter perform the necessary diagnosis for the runner. Additionally, the comparison between equation-derived measurements and actual measurements revealed no statistically significant difference. This confirms the possibility of using the equation to convert front view measurements to back view measurements despite having a low R^2 value. Future research could define reference norms and cut-off values based on front-view camera measurements of the foot inversion angle. This novel strategy might potentially be applied to all runners, regardless of the presence of crossover gait.

4.4 Limitations

This study used only one camera in each trial, positioned at different heights for the front and back views. The lack of synchronization between camera recordings may have compromised the accuracy of the results. The use of a single camera was done intentionally to simulate clinical settings where one camera view is often used to measure multiple kinematic variables, possibly due to the increased efficiency and limitations on manpower and equipment. For instance, the front camera view at the hip level could also measure knee adduction and contralateral hip drop (Dingenen et al., 2018a). Although this single-camera approach resulted in different gait cycles being analyzed and compared, it reflects the practical constraints encountered in clinical settings.

In addition, this study only measured 18 steps for each leg with varied gait patterns (e.g., crossover steps). Human locomotion such as running can present considerable variability between strides due to changes in motor control function (Stergiou et al., 2006). Previous studies that have attempted to identify the number of strides required for kinematic variables to reach a stable mean by performing a sequential estimation technique showed varying results. Dingenen et al. (2018a) reported a minimum of 7 steps while Oliveira and Pirscoveanu (2021) recommended a minimum of 25 steps were necessary to attain a stable kinematic mean. Therefore, although we have chosen a value within the higher range, it remains possible that the foot inversion angle measurements in this study did not reach a stable mean.

Another limitation is the use of a visual-based approach to identify the initial foot strike of running. This approach may be less accurate when compared to using an instrumented treadmill that provides ground reaction forces for quantitative identification (e.g., threshold of 20 N). From video recordings sampled at 60 Hz, Ghoussayni et al. (2004) found that the error in detecting the initial foot strike was less than 2 frames (0.033 s) when comparing to using a force platform. This present study used a high-speed video camera with 120 Hz recording frequency and hence a higher accuracy in initial foot strike identification can be expected.

5 Conclusion

This study explored the feasibility of measuring foot inversion angle at the initial foot strike of running from the front camera view. This method is particularly useful in crossover gait situations where the rearfoot is blocked in the back camera view. The findings indicate that the foot inversion angle at the initial foot strike is greater when measured from the front camera view than the back camera view. Researchers, practitioners, and clinicians are therefore cautioned against using measurements from different camera views interchangeably. A regression equation was also established to help clinicians translate the foot inversion angle measured from the front camera view to the standard rearfoot inversion angle measured from the back camera view. This will ensure accurate clinical diagnosis, using normative values that are currently derived from the back camera view.

Data availability statement

The raw data supporting the conclusion of this article will be made available by the authors, without undue reservation.

Ethics statement

The studies involving human participants were reviewed and approved by the Nanyang Technological University Institutional

Review Board (IRB-2021-124). The patients/participants provided their written informed consent to participate in this study.

Author contributions

PK secured funding and provided supervision. RL and PK contributed to conception and design of the study. RL, MI, and MH contributed to data collection. RL, JP, and PK were responsible for data organisation and management. RL and MI performed video analysis. MI and JP ran the statistical analysis. MI wrote the first draft of the manuscript. All authors contributed to the article and approved the submitted version.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Abnormal balance control mechanisms during dynamic reaching forward and quiet standing in patients with anterior cruciate ligament reconstruction

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Purpose: Postural instability and decreased balance control ability have been observed in patients after anterior cruciate ligament (ACL) reconstruction. Herein, we examined the abnormal balance control mechanisms of these patients during dynamic reaching forward and quiet standing, providing a quantitative index for rehabilitation assessment.

Methods: We enrolled ACL reconstruction patients 6–8 months after surgery, and 14 gender- and age-matched healthy volunteers. The IKDC and Lysholm were applied in each patient after ACL reconstruction. All participants conducted the quiet standing and reaching forward (RF) tests at the specified locations on force plates. The ground reaction force, center of pressure (COP), and kinematics signals were recorded. The maximal reach distance (MRD), speed of RF, length of COP, peak speed of COP in anterior-posterior direction (AP-COP), and weight bearing ratio (WBR) of the affected limb were calculated in the RF test. The COP speed, COP amplitude, frequency components of COP and WBR were extracted during quiet standing.

Results: We observed that the speed of RF in the patients after ACL reconstruction was significantly lower than that of controls ($p < 0.05$). The COP length during RF was positively correlated with the Lysholm scale in the affected limb of patients ($r = 0.604$, $p < 0.05$). The peak of AP-COP speed during RF in the affected limb of patients was significantly lower than that of the healthy controls ($p < 0.05$), and positively correlated with the IKDC scale ($r = 0.651$, $p < 0.05$). WBR on the affected limb of patients during RF were significantly lower than that of controls ($p < 0.05$). The mean ($r = -0.633$, $p < 0.05$) and peak ($r = -0.643$, $p < 0.05$) speeds of COP during quiet standing were negatively correlated with the IKDC scale value. The amplitude of AP-COP on the contralateral side of patients was significantly higher than that of controls during quiet standing ($p < 0.05$).

Conclusion: Patients after ACL reconstruction performed decreased postural control capacity, especially in dynamic balance, and were accompanied by deficiencies in proprioception. The COP length, peak speed of COP during RF and COP speed during quiet standing could be considered as quantitative index of balance function assessment after ACL reconstruction.

KEYWORDS

ACL reconstruction, balance control mechanism, center of pressure, reaching forward, rehabilitation

1 Introduction

Stable posture control for quiet standing and movement is a fundamental motor function during daily activities. During convalescence following anterior cruciate ligament (ACL) reconstruction, patients performed imperfect motor function, such as postural instability, abnormal gait, and difficulty in sit-stand transfer (Clark et al., 2014; Difabio et al., 2018; Hoch et al., 2018). The balance control capacity is the foundation of various motor functions. Evaluating the postural stability for balance control during dynamic and static conditions may provide quantifications for muscular disorders and would aid in the development of strategies to improve the balance capacity of patients with ACL reconstruction during rehabilitation (Kim et al., 2022).

Compared to these technologies for postural stability analysis, such as subjective assessment, limb vibrations, trunk force line, and spatial kinematics, the center of pressure (COP) is relatively easy to record, is linked to postural control, and indicates critical information regarding the muscular coordination in lower-limb (Ty Hopkins, 2014; Kilby et al., 2016). In a previous study, patients with ACL reconstruction showed increased amplitude variability and speed value of COP compared to healthy individuals, reflecting a decreased balance control capacity and aggravated postural sway during standing (Clark et al., 2014).

The reaching forward (RF) test during standing has previously been applied to evaluate balance capacity clinically. A previous study reported that the measurement of the maximal distance during reaching forward horizontally with arms outstretching can be considered as a reliable parameter to evaluate the balance

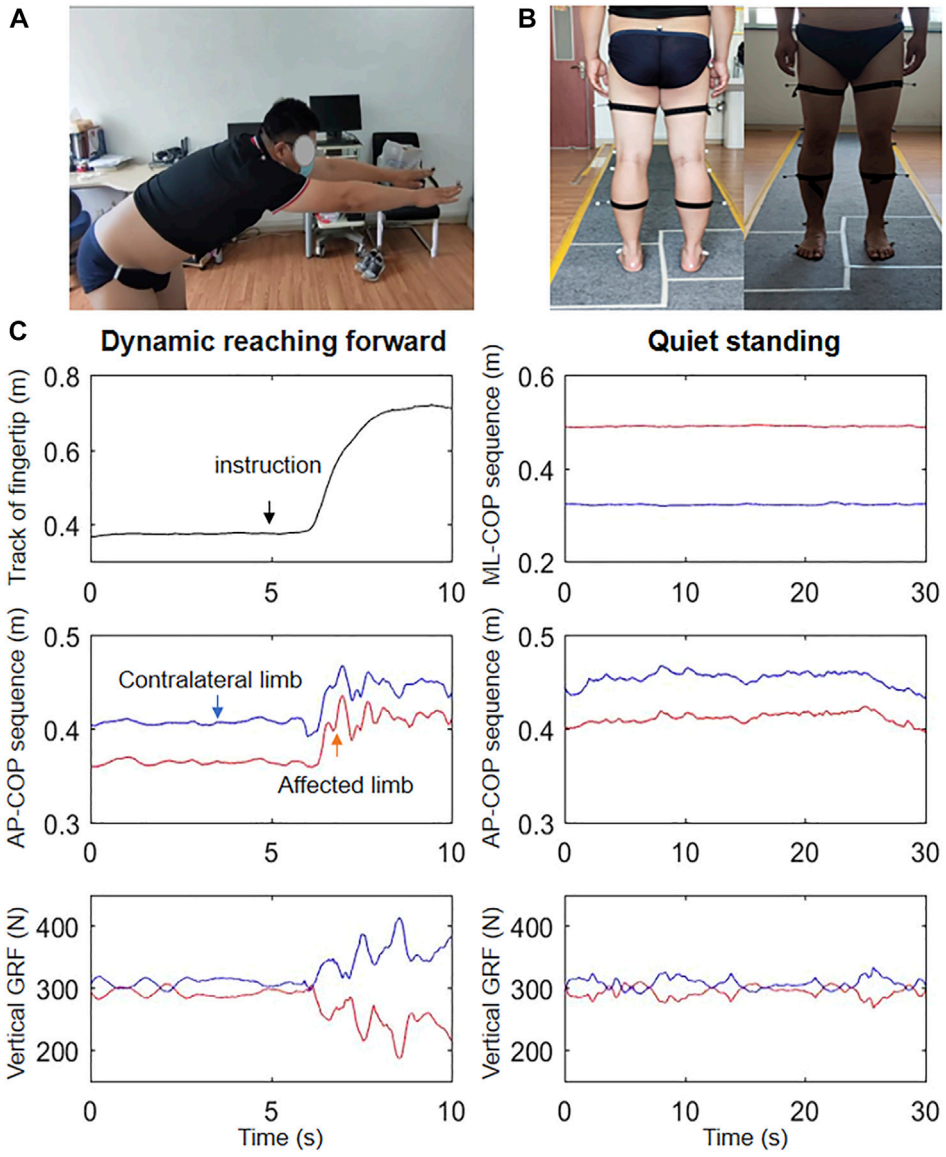

FIGURE 1
Overview of the **(A)** reaching forward and **(B)** quiet standing tasks performed during the experiment, and **(C)** the tracks of the fingertips, AP-COP, and vertical RFG signals during reaching forward and the ML-COP, AP-COP, and vertical RFG signals during quiet standing in a representative patient.

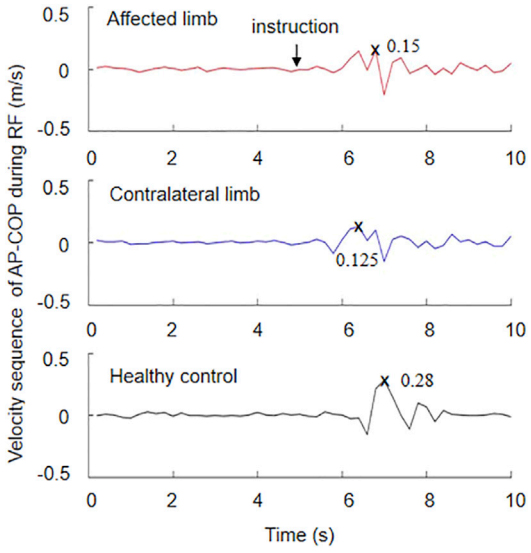

FIGURE 2
Velocity sequence of AP-COP during reaching forward calculated by a 200 ms slide-window.

capacity in dynamic condition (Bohannon et al., 2020). In the RF test, greater maximal reach distance (MRD) indicates larger stability boundary and lower risk of falls in daily movement. Indeed, it has been reported in previous study that patients with low back pain showed decreased MRD and muscle flexibility during RF (Johnson & Thomas, 2010). The COP antedisplacement can be used to indicate the security boundary during postural adjustment and reflects the movement of muscles to control the rotation of ankle (Gandolfi et al., 2021).

The insufficient balance function could be observed in patients with ACL reconstruction. However, it remains unknown that the changes of COP characteristics and the relationship between these characteristics and knee function during RF 6 months after ACL reconstruction. This study aimed to investigate the dynamic and static balance control capacity of patients with ACL reconstruction. We therefore determined whether patients with ACL reconstruction show balance disorders by exploring the MRD values, COP antedisplacement, peak speed of COP in anterior-posterior direction (AP-COP) during RF, amplitude variability of COP, and speed of COP trajectory during quiet standing. This study explored the abnormal postural control pattern and could provide quantitative assessment indexes for rehabilitation after ACL reconstruction.

2 Methods

2.1 Participants

We enrolled 14 patients who had undergone ACL reconstruction 6–8 months prior and formal rehabilitation training, as well as 14 healthy volunteers as controls. The sample size was determined by power analysis. Patients were recruited from the Qilu Hospital of Shandong University, from 14 August 2021 to 21 June 2022. All patients (aged 18–60 years old) had been clinically diagnosed with unilateral ACL injury, had undergone surgery 6–8 months prior, and could complete all tasks in the experiment independently. Individuals with a history of cardiovascular and cerebrovascular diseases, muscle or bone injuries in their low back or lower extremities, vertigo or vestibular disease, severe visual defects, or cognitive difficulties were excluded. The operator explained the purpose and potential risks of the experiment in detail to every subject. All subjects signed an informed consent form. The protocols of the current study had been approved by the medical ethics committee of Qilu Hospital of Shandong University (KYLL-202107-041). The study was conducted in accordance with relevant regulations and guidelines.

2.2 Experimental set-up

A three-dimensional motion and capture system (BTS Bioengineering Corp., Italy) with six cameras and two force plates was used during the experiment. The three-dimensional positions of four retro-reflective markers placed on the bilateral shoulders and tips of the middle fingers were recorded at a sampling frequency of 100 Hz (Figure 1A). Two adjacent force plates, spliced together and embedded horizontally into the ground, measured the COP and ground reaction force (GRF) signals at a sampling frequency of 200 Hz. The dynamic signals were synchronous with the kinematic signals.

2.3 Test protocol

The IKDC and Lysholm scale values of each patient after ACL reconstruction were acquired by the same trained personnel before testing. For the first task, all subjects conducted standing upright at the specified locations of the two force plates (each foot on a different plate) for 30 s (Figure 1B). For the second task, all subjects stood upright at the identical locations in the first task, while maintaining the bilateral arms and hands outstretched horizontally. The operator gave the order. Subjects reached forward as far as capable without lifting up their heels (Figure 1A). Subjects repeated this process using the right arm, and both arms simultaneously. Two trails were conducted for each task with 2-min interval between trials. Participants were asked to practice the movements several times before measurements.

2.4 Data analysis

Figure 1C depicts the fingertip tracking, AP-COP sequence, and vertical GRF during dynamic forward, as well as the COP in the medial-lateral direction (ML-COP) sequence, AP-COP sequence, and vertical RFG during quiet standing. The subject stood on the force plates stably before the system triggered. The command 'reach forward' was made at the 5th second of standing upright with the bilateral arms and hands outstretched horizontally. The MRD was calculated as the difference between the maximal position and the preparation position, i.e., the mean value during the 3–4 s. The speed of forward movement was calculated as the MRD divided by time

TABLE 1 Descriptive characteristics of subjects.

Number		ACL reconstruction patients (n = 14)						Control (n = 14)			
	Gender	Age(years)	Affected side	Height(cm)	Weight(kg)	IKDC	Lysholm	Gender	Age(years)	Height(cm)	Weight(kg)
1	M	30	R	170	69	-	-	M	31	170	80
2	F	19	L	185	85	94.25	96	F	21	168	54
3	F	22	L	178	73	-	-	F	21	164	52.5
4	M	29	R	186	70	70.11	75	M	30	170	87
5	F	29	L	161	68.5	70.11	95	F	29	160	60
6	M	19	R	175	88	77.01	69	M	19	181	75
7	F	32	R	165	62	62.07	59	F	31	162	58
8	M	18	L	175	75	48.28	23	M	19	168	60
9	F	32	L	161	51	39.08	58	F	32	167	60
10	M	41	L	181	83	95.4	91	M	39	168	64
11	M	26	R	173	110	95.4	65	M	24	180	85
12	M	24	R	193	80	90.8	100	M	23	172	65
13	M	40	R	168	63	81.61	89	M	39	170	63
14	M	19	L	183	92	85.06	89	M	21	175	95

The n is the number of subjects.

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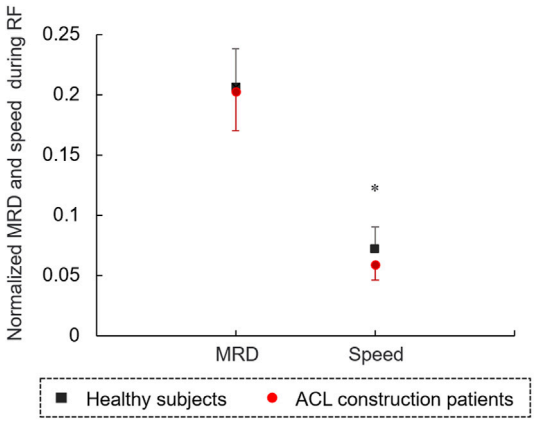

FIGURE 3
The MRD and speed of forward movement, normalized by height during reaching forward.

during reaching forward. The maximal position was defined the peak point during reaching forward. The length of COP was calculated as the difference between the maximum AP-COP and the minimum AP-COP during RF. These parameters were normalized to height. The velocity sequence of AP-COP during RF was calculated by a 200 ms slide-window (Figure 2). The peak speed was the maximum value of the velocity sequence. The weight bearing ratio (WBR) on the affected side was defined as the GRF of the affected limb divided by the total GRF. The MATLAB 2016a (The Mathworks, Natick, MA, USA) was used for the data analysis.

The amplitude of COP variability was quantified using the formula for standard deviation (SD), as follows:

$$SD = \sqrt{\frac{1}{n}\sum_{i=1}^{n}(x(i)-\bar{x})^2} \quad (1)$$

where $x(i)$ is the magnitude of the COP signal at each time point i, $\bar{x}$ is the mean value of the $x(i)$, and n is the signal length of $x(i)$.

The velocity sequence (V) of COP during quiet standing was defined by the equation:

$$V(i) = \frac{\sqrt{(x(i+1)-x(i))^2 + (y(i+1)-y(i))^2}}{\Delta t} \quad (2)$$

where $x(i)$ and $y(i)$ are the magnitude of the AP-COP and ML-COP signals at each time point i, Δt is the time between two adjacent sampling points. The mean and peak speeds of COP during quiet standing was the mean and peak values of the COP velocity sequence.

The power spectral density (PSD) of the COP sequence was calculated to analyze the low-frequency (LF: less than 1 Hz) and high-frequency (HF: 1–10 Hz) components of COP signal. The COP had been filtered with a cutoff frequency of 15 Hz by a Butterworth low-pass filter.

Statistical analyses were performed by SPSS 20.0 (SPSS Inc., Chicago, IL). All variables were verified as normal distribution by the Kolmogorov-Smirnov test. Independent t-tests were applied to examine the differences between patient group and control group, patients' affected limbs and healthy limbs, patients' contralateral limbs and healthy limbs. The healthy limbs of the controls were corresponding to the affected limbs of the paired patients. For example, the patient's affected limb is left limb, then the left limb of the healthy subject was selected as the healthy limb. Paired t-tests were used to examine the difference between the affected limb and the contralateral limb. No interaction was considered across these t-tests. Similarly, bivariate correlation analysis was performed to assess the correlations between the IKDC/Lysholm scales and variables during RF and quiet standing. Values were considered statistically significant at $p < 0.05$.

3 Results

The characteristics of the 14 patients with ACL reconstruction (5 women and 9 men; aged 27.74 ± 7.27 years) and 14 healthy controls (5 women and 9 men; aged 27.07 ± 6.65 years) are shown in Table 1. There was no significant difference in age (t = 0.026, p =

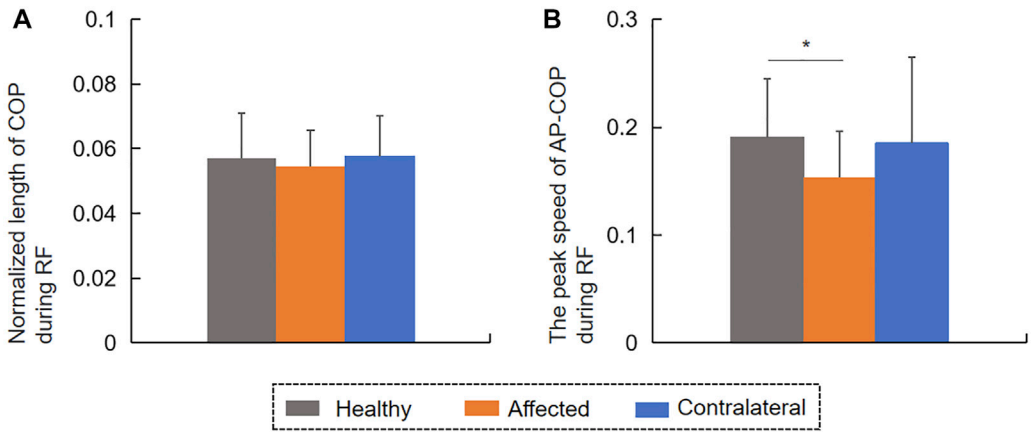

FIGURE 4
The **(A)** COP length normalized by height and the **(B)** peak speed of AP-COP during reaching forward.

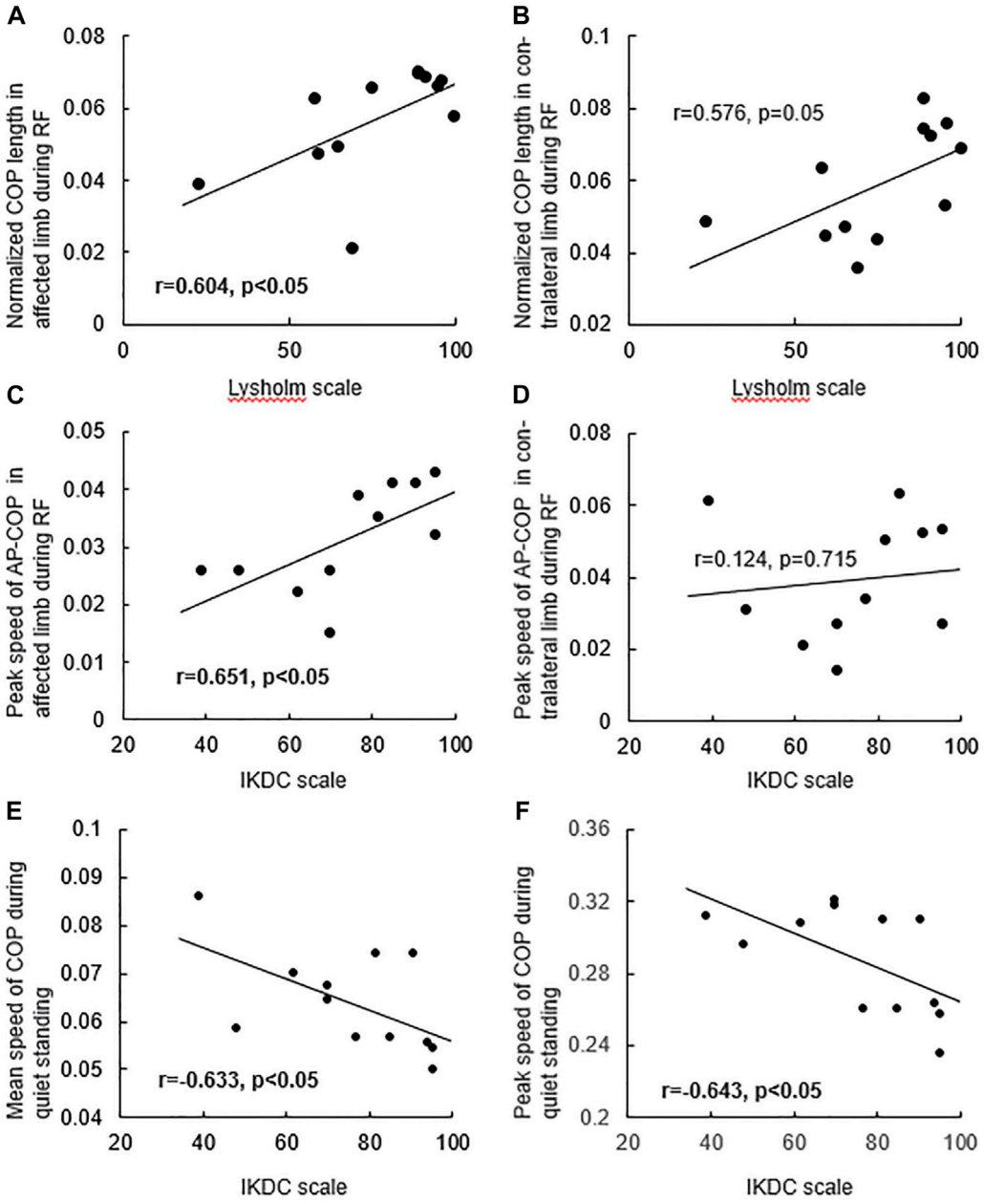

FIGURE 5
The relationship between the COP length normalized by height in **(A)** affected limb, **(B)** contralateral limb and Lysholm scale during reaching forward, and the relationship between the peak speed of AP-COP in **(C)** affected limb, **(D)** contralateral limb and IKDC scale during reaching forward, and the relationship between the **(E)** mean speed, **(F)** peak speed of COP and IKDC scale during quiet standing.

0.979), height (t = 1.485, $p = 0.150$) and weight (t = 1.851, $p = 0.078$) between two groups. The results of the MRD and speed normalized by height during reaching forward are shown in Figure 3. The speed of RF in the patients with ACL reconstruction normalized by height was significantly lower than that of controls (t = −2.983, $p < 0.05$). No significant difference was observed in the MRD value between the patients with ACL reconstruction and controls ($p > 0.05$).

No significant difference was found in the COP length normalized by height between affected and healthy limbs, or between contralateral and healthy limbs, or between affected and contralateral limbs during RF ($p > 0.05$, Figure 4A). The peak speed of the AP-COP in the patients' affected limb was significantly lower than that of healthy limb during RF (t = −2.279, $p < 0.05$, Figure 4B). The normalized COP length in the patients with ACL reconstruction was positively correlated with the Lysholm scale in affected limb (r = 0.604, $p < 0.05$, Figure 5A), instead of the contralateral limb (r = 0.576, $p = 0.05$) (Figure 5B) during RF. The peak speed of AP-COP in the patients' affected limb was positively correlated with the IKDC scale (r = 0.651, $p < 0.05$) (Figure 5C); however, no significant

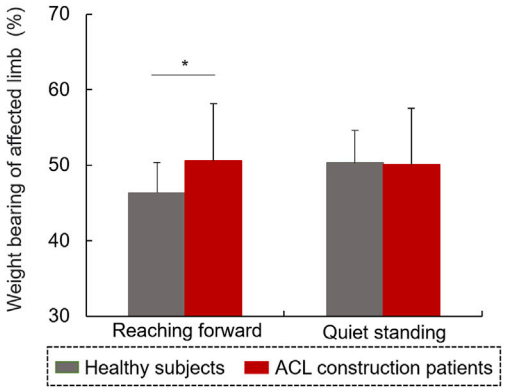

FIGURE 6
Weight bearing of the affected limb during the reaching forward task and static standing.

correlation was found between AP-COP and the IKDC scale in the contralateral limb (r = 0.124, p = 0.715).

The WBR on the affected side in patients with ACL reconstruction was significantly lower than that (50.61%) in the controls during RF (t = −2.17, p < 0.05, Figure 6). There was no significant difference in the WBR on the affected limb between the two groups during RF (p > 0.05).

No significant differences were observed in terms of mean or peak speed between patients with ACL reconstruction and controls during quiet standing (Figure 7A). The amplitude of AP-COP variability in contralateral limb was significantly higher than that in healthy limb during standing (t = 2.196, p < 0.05, Figure 7B). No significant differences were found in the amplitude of ML-COP variability between affected and healthy limbs, or between contralateral and healthy limbs, or between affected and contralateral limbs. The mean (r = −0.633, p < 0.05, Figure 5E) and peak speed (r = −0.643, p < 0.05, Figure 5F) values were negatively correlated with the IKDC scale values.

The frequency components of COP signals during quiet standing were shown in Figure 8. The HF component of AP-COP in patients with ACL reconstruction was significantly higher than that of healthy subjects (t = 2.074, p < 0.05, Figure 8A). There was no significant difference in HF component of ML-COP, LF component of AP-COP or ML-COP between patients and healthy subjects.

4 Discussion

This study investigated the balance control capacity during static and dynamic conditions of patients after ACL reconstruction. The patients with ACL reconstruction showed a lower speed of RF compared with controls, although no difference was observed in the MRD values between the two groups. A reduced peak speed of the affected AP-COP indicates an inadequate control ability of COP transfer during dynamic balance condition in patients with ACL construction. In our study, the patients' affected limbs bore less weight only during dynamic RF instead of quiet standing. Patients with ACL reconstruction further showed a higher amplitude of COP variability and HF component of COP in the AP direction compared with controls during standing. In the dynamic RF condition, COP antedisplacement value was positively correlated with the Lysholm scale value. In the static standing condition, the values of mean and peak speeds of COP were negatively correlated with the values of IKDC scale. Overall, all of these results contribute to our understanding of patients' balance function 6 months after ACL reconstruction, and will help in formulating targeted strategies for rehabilitation training in clinical settings.

COP amplitude variability quantified by SD is considered to be an indicator of balance control and postural stability (Rakhra & Singer, 2022). The increased amplitude variability of AP-COP in the contralateral side reflects compensation of the contralateral limb and enhancive postural sway after ACL reconstruction during quiet standing (Yu et al., 2008). It has been reported that the LF and HF components of COP reflect the sensitivity of vestibular and visual information, and

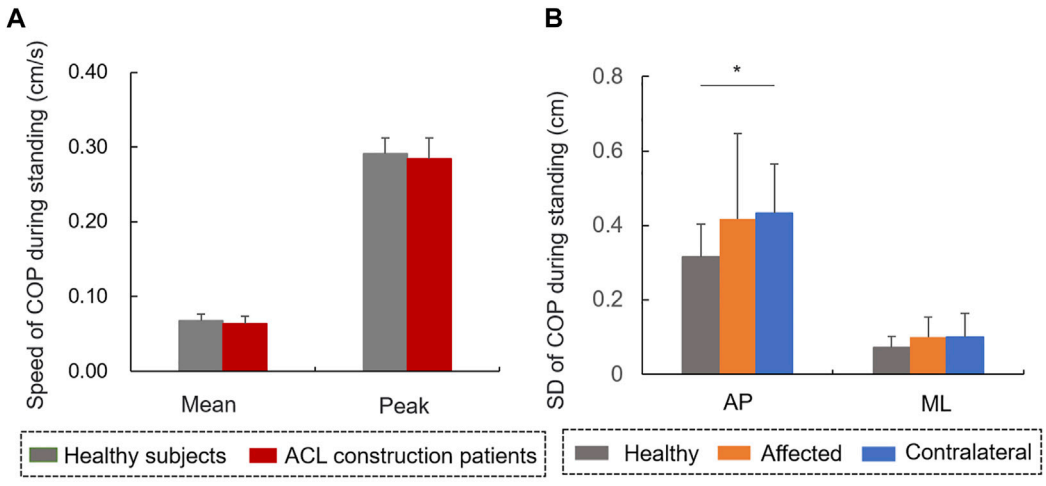

FIGURE 7
The **(A)** speed of COP trajectory and **(B)** amplitude of COP during quiet standing.

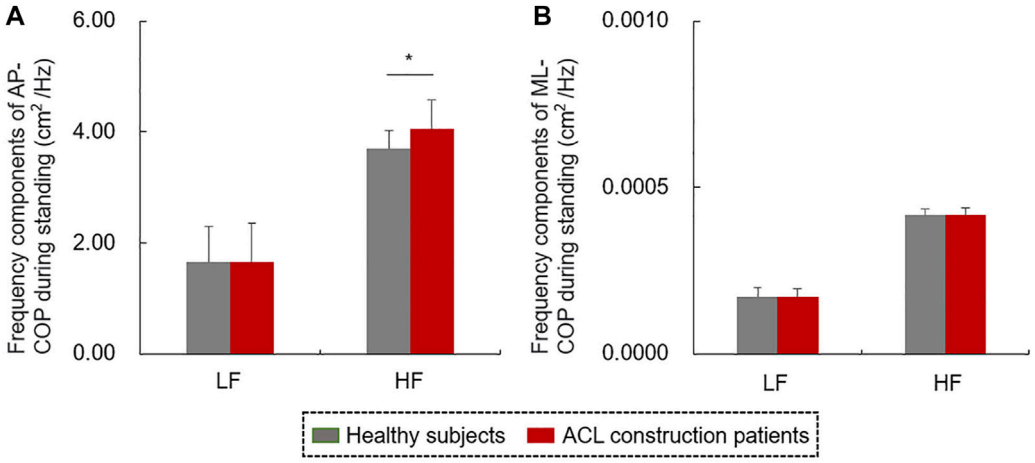

FIGURE 8
The frequency components of **(A)** AP-COP and **(B)** ML-COP signals during quiet standing.

somatosensory inputs including proprioception (Kouzaki & Masani, 2012). Increased HF component of AP-COP indicated the reduced proprioception in patients with ACL reconstruction (Mcchesney & Woollacott, 2000; Kouzaki & Masani, 2012). No similar results were found in ML-COP variability, which indicates that the balance control in the AP, but not the ML direction, was influenced by ACL injury. In addition, no significant differences were found in terms of mean or peak speed of COP trajectory, or the WBR of affected limb between patients with ACL-reconstruction and controls during quiet standing. These findings suggested that patients had partly regained their static balance control ability 6 months after ACL reconstruction. It has been reported that a higher COP speed indicates increased postural instability and falling risk (Yilmaz & Bulbuloglu, 2021). In the current study, both the mean and peak speed values of COP trajectory during quiet standing were negatively correlated with the IKDC scale values, indicating that the decrease of knee function after ACL injury, including pain, stiffness, and swelling, causes instability of the static balance control and fast postural adjustment (Bodkin et al., 2018).

The MRD, defined as the distance the fingertip travels while RF, indicates the balance boundary and postural control capacity of the participant (Lin et al., 2010). In the current study, the MRD value in patients with ACL was not different from that of controls, demonstrating that patients had a normal reaching boundary 6 months after ACL reconstruction. However, the reduced speed of RF suggested that these patients still required more time to reach the target distance. This inefficient strategy may expend more energy and could be linked to a lack of confidence and fear of falling.

The lower weight bearing on the affected limb indicated inadequate supporting capacity requiring compensation by the contralateral limb during dynamic balance control after ACL reconstruction. Previous studies have reported that altered nerve reflex pathways may partially inhibit α motoneuron excitability of the quadriceps and increase hamstring muscle activity in patients after ACL reconstruction (Burland et al., 2020; Suh et al., 2021). Additionally, weight bearing is known to be related to quadriceps muscle strength and function in patients with ACL-reconstruction (Chmielewski et al., 2002). Therefore, this finding further implies that the inadequate supporting capacity was possibly contributed by the knee instability, pain, muscle strength weakness, and decreased muscle coordination in 6 months after ACL reconstruction.

COP signals during postural control provide neuromuscular and somatosensory information. A previous study demonstrated that muscle strength and proprioception are significantly associated with postural stability, as represented by COP variables, in patients with unilateral knee osteoarthritis (Zeng et al., 2022). In the present study, the reduced peak speed of AP-COP of the affected limb during RF suggested that ACL reconstruction reduced the control ability of COP transfer and changed the dynamic balance control strategy (Bulow et al., 2021). In addition, the peak speed of AP-COP of the affected limb positively correlated with the IKDC scale. These findings suggested that deficient knee function limited the forward transfer speed of COP. Although the COP antedisplacement of patients with ACL reconstruction did not differ from that of controls during RF, the antedisplacement value of COP positively correlated with the Lysholm scale value. This result implies that the narrowed equilibrium boundary during RF may be partly generated by knee instability, pain, and lock following ACL reconstruction. The peak speed of AP-COP during RF could be considered as an evaluation indicator for patients following ACL construction in clinical practice. In clinical rehabilitation, balance control and proprioception training, such as vertical vibration and single-leg standing with eyes closed, supporting capacity training of affected limb should be considered to improve balance function (Faggal et al., 2019).

This study had some limitations. First, the sample size was too small to fully reveal the relationship between the ACL reconstruction and balance control ability. Second, we did not record surface EMG signals, or behavioral, and clinical outcomes. Further studies are needed to address these gaps. Third, this is a single center study.

5 Conclusion

At 6 months after surgery, patients with ACL reconstruction regained part of their balance ability, but showed decreased postural control and loading ability of the affected limb, especially in terms of

dynamic balance, suggesting the adoption of an abnormal postural control strategy. The COP length, peak speed during RF and COP speed during quiet standing could be considered as quantitative assessment index for balance function of patients with ACL reconstruction. Clinicians should be aware of the dynamic balance training, correcting postural control strategy, and improving supporting capacity in affected limb should be a primary goal during rehabilitation following ACL reconstruction.

Data availability statement

The original contributions presented in the study are included in the article/Supplementary Material, further inquiries can be directed to the corresponding author.

Ethics statement

The studies involving human participants were reviewed and approved by the medical ethics committee of Qilu Hospital of Shandong University (KYLL-202107-041). The patients/participants provided their written informed consent to participate in this study.

Author contributions

Conceptualization: HW and WW. Data curation: WW. Formal analysis: HW and WW. Funding acquisition: HW. Investigation: RS, XL, and XR. Methodology: HW and WW. Project administration: HW. Resources: HW. Software: WW and KZ. Supervision: HW. Validation: HW and WW. Visualization: HW and WW. Writing—original draft: WW. Writing—review and editing: HW and WW. All authors contributed to the article and approved the submitted version.

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Conflict of interest

The authors declare that the research was conducted in the absence of any commercial or financial relationships that could be construed as a potential conflict of interest.

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Supplementary material

The Supplementary Material for this article can be found online at: https://www.frontiersin.org/articles/10.3389/fphys.2023.1176222/full#supplementary-material

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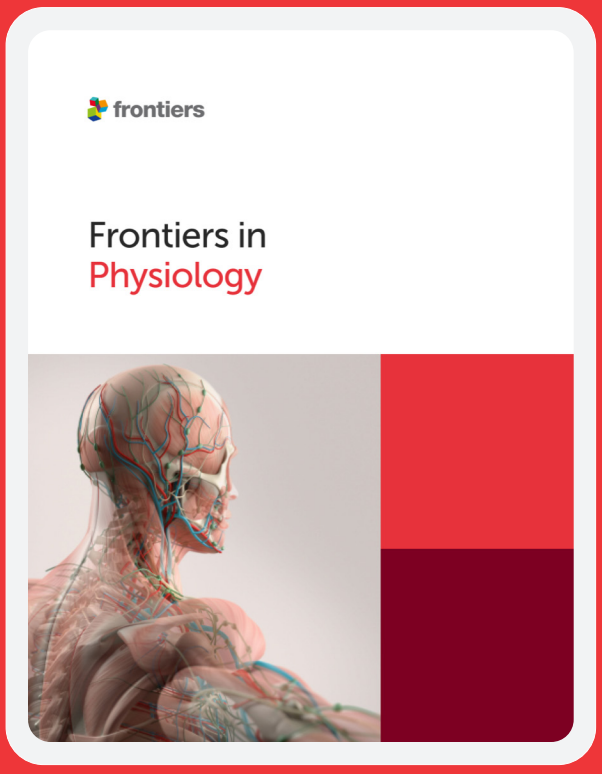

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