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

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

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Editorial: Biomechanical performance and relevant mechanism of physical medicine and rehabilitation for neuromusculoskeletal disorders, volume II: aiding injury prevention and improving rehabilitation with a better understanding of relevant biomechanical mechanisms

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Editorial on the Research Topic
Biomechanical performance and relevant mechanism of physical medicine and rehabilitation for neuromusculoskeletal disorders, volume II: aiding injury prevention and improving rehabilitation with a better understanding of relevant biomechanical mechanisms

Introduction

Neuromusculoskeletal disorders, such as sports-related injuries, e.g., chronic ankle instability (CAI), anterior cruciate ligament (ACL) injuries or patellofemoral pain (PFP); neurological disorders, e.g., cerebral palsy or stroke; and chronic disorders, e.g., knee osteoarthritis (KOA) or chronic obstructive pulmonary disease (COPD); are debilitating conditions that can lead to substantial functional impairment or even permanent disability. Despite its significant impact and ongoing burden at both personal and societal levels, the underlying mechanisms of physical medicine and rehabilitation remain poorly understood. The limited understanding in this field may impede the development of clinical rehabilitation strategies.

Biomechanical assessment may help better understand the mechanisms underlying physical medicine's effectiveness and rehabilitation for neuromusculoskeletal disorders. From advanced infrared motion capture systems to finite element models, various modalities are available to investigate the changes in the physical function of individuals with neuromusculoskeletal disorders and to examine the differences in their physical function compared to uninjured individuals. Therefore, this Research Topic aimed to gather recent advancements to enhance the understanding of biomechanical performance and relevant mechanisms of physical medicine and rehabilitation for neuromusculoskeletal disorders. Sixteen articles have passed the peer review and were finally published. Among these papers, 10 works explained sports-related injuries, four articles explored neurological disorders, and two works discussed chronic disorders.

Mechanisms of sports-related injuries

Preventing sports-related lower limb injuries remains a key area of research interest, with the primary objective of minimizing injury risk and elucidating the underlying mechanisms. Zou and their group investigate the effects of fatigue and anticipation on biomechanical risk factors of ACL injury during 180° pivot turns in female soccer players (Zou et al., 2024). They observed that fatigue and unanticipated tasks negatively impacted the lower-limb biomechanics and diminished movement performance among female soccer players, thereby increasing the risk of ACL injury during pivot turns. The authors recommended that cognitive training should be strengthened in female soccer players to mitigate the risk of non-contact ACL injuries under unanticipated scenarios during competitions. Xue and coworkers recruited 50 male participants, including those who had undergone ACL reconstruction on either the dominant (ACLR-D) or nondominant (ACLR-ND) limb, as well as healthy controls, and compared the biomechanical characteristics of bilateral limbs among these groups during a drop vertical jump task (Xue et al.). It was observed that compared to the surgical limbs, the nonsurgical limbs of ACLR patients may suffer an increased risk of ACL injury due to altered landing mechanics and neuromuscular control strategies. Moreover, the nonsurgical limb in the ACLR-ND group may be at greater risk than that in the ACLR-D group. These reports indicate that the impact of limb dominance should be considered in the rehabilitation of ACLR patients for better return to sport.

Within sports-related injuries, PFP is also one of the most prevalent conditions and differs from ACL injuries in that it is a nontraumatic knee problem. Nagahori and Ho focused on the relationship between patellar cartilage thickness and knee external rotation during squatting in individuals with and without PFP (Nagahori and Ho). They observed that thinner patellar cartilage was associated with increased knee external rotation during bilateral squatting, regardless of PFP status. These observations suggested that increased knee external rotation during squatting may negatively affect patellar cartilage thickness. The authors recommended that clinicians carefully assess KER during squatting in patients with or at risk of PFP. In addition to alterations in lower extremity kinematics, central nervous system adaptations have also been observed in individuals with PFP. Ho and colleagues investigated the central activation ratio of the gluteal muscles in individuals with and without PFP (Ho et al.). Their results suggested a tendency for individuals with PFP to exhibit a lower gluteus maximus central activation ratio, although the difference was not statistically significant. Furthermore, a higher gluteal central activation ratio was associated with better function in this population. The above studies provide valuable insights for optimizing rehabilitation strategies in individuals with PFP.

Investigating the biomechanical characteristics of individuals with CAI during landing from a height and strategies for reducing the potential risk of ankle sprains remains a primary focus in rehabilitation research. Compared to individuals without CAI, Zhong and colleagues reported that those with CAI demonstrate a greater potential risk of injury, as indicated by greater ankle inversion angles and angular velocities during drop landings (Zhong et al.). Their study further revealed that, compared to single-task conditions, the ankle inversion angles and angular velocities under dual-task conditions differed among individuals without CAI. In contrast, no significant changes were observed in individuals with CAI (Zhong et al.). These reports suggest that individuals with CAI may exhibit maladaptive neuroplastic changes at the spinal and cortical levels, which could increase their potential risk of injury under dual-task conditions. Wang's group investigated the biomechanical characteristics of the lower limbs during single-leg drop landings in individuals with unilateral functional ankle instability under various attention focus strategies (Wang et al.). They observed that external focus strategies promote a conservative landing strategy, while internal focus may enhance lower limb stability. Integrating these strategies into functional ankle instability rehabilitation programs can help reduce the risk of reinjury. The finite element method, by simulating the mechanical responses of the ankle's lateral ligament, provides an effective means to evaluate the risk of ankle injuries (Zhu et al., 2025). Zhou et al. employed a finite element method to investigate alterations in neural control and stress response distribution during landing in patients with ankle ligament injuries (Zhou et al.)The results showed that the anterior talofibular and calcaneofibular ligaments' laxity increases stress on the metatarsals. This ligament laxity may also trigger muscle compensation, which can further affect ankle joint stability and increase the risk of ankle injury.

The biomechanical changes in the lower extremity caused by CAI are not limited to the ankle joint but also affect the proximal knee and hip joints, thereby increasing the risk of injury (Zhang et al., 2024). Based on retrospective studies, He et al. reported that patients with CAI exhibit changes in biomechanical parameters during landing that are associated with an increased risk of ACL injury (He et al.). These changes include increased hip and knee extension moments, reduced hip flexion angles, elevated peak vertical ground reaction forces, and greater trunk lateral flexion angles. These reports provide a valuable basis for designing targeted prevention measures and rehabilitation programs to help reduce the risk of ACL injuries in individuals with CAI.

Running is a common form of exercise, but improper technique may contribute to sports-related injuries. Verdel et al. recruited 15 injury-free female runners to examine the effects of running speed, incline, and fatigue on the calcaneus angle (Verdel et al.). They observed that higher running speeds and fatigue conditions may elevate injury risk by increasing the range of motion. In contrast, inclined running may reduce this risk by limiting excessive eversion and range of motion. These reports provide useful insights for developing injury prevention strategies relevant to runners and researchers.

Rehabilitation strategies for sports-related injuries

Musculoskeletal injuries can lead to significant functional deficits mediated by the central nervous system (CNS), including increased arthrogenic muscle inhibition and altered neuroplasticity (Dong et al., 2024). In sports-related injury rehabilitation, addressing CNS dysfunction is essential for optimal recovery. Huang et al. utilized transcranial direct current stimulation (tDCS) and Bosu intervention on the injury potential during drop landing in people with CAI (Huang et al.). They reported that, compared to the Bosu intervention alone, transcranial direct current stimulation combined with the Bosu intervention was more effective in reducing peak ankle inversion angular velocity, plantarflexion angle at the moment of peak ankle inversion, and advancing time to peak ankle inversion, and that both interventions reduced the peak ankle inversion angle. Their reports provide new insights into the clinical development of rehabilitation strategies for individuals with CAI. They indicate that integrating CNS-directed interventions, such as transcranial direct current stimulation, into conventional functional training can help reduce the risk of recurrent ankle injuries.

Mechanisms and rehabilitation strategies of neurological disorders

A common neurodevelopmental disorder, Cerebral Palsy (CP), leads to significant and lasting impairments of the neuromuscular system. Pontiff et al. investigated the relationships of the Power Leg Press performance with walking capacity and self-reported performance and participation in ambulatory individuals with CP to understand how muscle power influences activity capacity and participation (Pontiff et al.). Their observations demonstrated a significant positive association between leg press power, walking capacity, and self-reported walking performance and mobility-related participation in ambulatory individuals with CP. These results suggest that clinicians should incorporate lower extremity power into their assessments of individuals with CP and recommend using the Power Leg Press test. In another work, 10 participants with typical development and 8 participants with CP were included (Damiano et al.). The results showed that after the body weight supported treadmill training, the CP group exhibited beneficial effects on kinematics, which supports the basic premise for applying such interventions in neurorehabilitation at the body structure level.

Neurodegenerative disorders are emerging as major public health challenges. Stroke is one of the leading causes of adult death worldwide and can cause severe movement dysfunction and limitations, such as walking difficulties. Liu et al. explored immediate changes in temporal and spatial parameters of gait and the joint angles in people suffering from stroke throughout the entire gait cycle after the application of the lower extremity elastic strap binding technique (Liu et al.). This report stated that for people suffering from stroke, the lower extremity elastic strap binding technique can help reduce the hip and knee flexion limitations and decrease the ankle plantarflexion and inversion angles during walking. These benefits potentially result from proprioceptive feedback that stimulates changes in the excitability of the motor cortex to promote effective coordinated movement (Song et al., 2021). This offers a complementary rehabilitation strategy for improving gait in people who have suffered a stroke with foot drop and limited hip and knee flexion.

In another study, Yang et al. recruited ten individuals with stroke to assess spatiotemporal gait parameters and symmetries immediately after split-belt treadmill training and single-belt treadmill training, as well as after a 5-minute rest following each intervention (Yang et al.). The study concluded that single-belt and split-belt treadmill training effectively improved gait speed and step length on the shorter side in patients with stroke. Furthermore, the results showed improvement in step length symmetry immediately after split-belt treadmill training, without impairing other temporal symmetries. However, this effect diminished after a 5-minute rest. This work highlights the potential of split-belt treadmill training to enhance gait symmetry. It may provide valuable insights for developing more efficient and safer gait rehabilitation strategies for patients with stroke.

Rehabilitation strategies for chronic disorders

The pain associated with KOA has increased the likelihood of tripping over obstacles. Pain-relieving transcutaneous electrical nerve stimulation (TENS) is widely utilized in treating KOA; however, it exhibits limited analgesic effects and modest benefits for functional ability. On that basis, a total of 23 participants with KOA were randomized to either the tDCS + TENS group or the TENS-only group to evaluate the effects of the two interventions on pain reduction and gait optimization during obstacle crossing among older adults with KOA (Zhang et al.). The study concluded that the combination of tDCS and TENS was significantly more effective than TENS alone in reducing pain and improving gait adaptability during obstacle negotiation among older adults with KOA. These reports offer new perspectives to guide the development of future rehabilitation strategies for individuals with knee osteoarthritis.

Among older adults, COPD is prevalent and exerts substantial impacts on quality of life, morbidity, and mortality. Vocal therapy enhances respiratory muscle strength and endurance, with singing training emerging as an increasingly popular pulmonary rehabilitation program for patients with COPD. Qiao et al. investigated the content of vocalization training for patients with COPD by examining differences in respiratory muscle activation across various vocalization tasks (Qiao et al.). The study observed that vocal loudness, rather than pitch or vowels, should be the key factor in voice training for these patients. This research provides clinical significance for implementing voice training.

This Research Topic presents the latest approaches to physical medicine and rehabilitation interventions for neuromusculoskeletal disorders. It encourages the use of biomechanical approaches to evaluate the effectiveness and explore the mechanisms of these interventions. We hope this work will inspire scholars researching biomechanical performance and relevant mechanisms of physical medicine and rehabilitation for neuromusculoskeletal disorders to explore these research frontiers further.

Author contributions

QW: Writing – review and editing, Writing – original draft. QS: Writing – review and editing, Writing – original draft, Conceptualization. PK: Writing – review and editing, Writing – original draft. CZ: Writing – review and editing, Writing – original draft. DW: Writing – original draft, Writing – review and editing. LL: Conceptualization, Writing – original draft, Writing – review and editing.

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Analysis of stress response distribution in patients with lateral ankle ligament injuries: a study of neural control strategies utilizing predictive computing models

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Background: Ankle sprains are prevalent in sports, often causing complex injuries to the lateral ligaments. Among these, anterior talofibular ligament (ATFL) injuries constitute 85%, and calcaneofibular ligament (CFL) injuries comprise 35%. Despite conservative treatment, some ankle sprain patients develop chronic lateral ankle instability (CLAI). Thus, this study aimed to investigate stress response and neural control alterations during landing in lateral ankle ligament injury patients.

Method: This study recruited twenty individuals from a Healthy group and twenty CLAI patients performed a landing task using relevant instruments to collect biomechanical data. The study constructed a finite element (FE) foot model to examine stress responses in the presence of laxity of the lateral ankle ligaments. The lateral ankle ligament was modeled as a hyperelastic composite structure with a refined representation of collagen bundles and ligament laxity was simulated by adjusting material parameters. Finally, the validity of the finite element model is verified by a high-speed dual fluoroscopic imaging system (DFIS).

Result: CLAI patients exhibited earlier Vastus medialis ($p < 0.001$) and tibialis anterior ($p < 0.001$) muscle activation during landing. The FE analysis revealed that with laxity in the ATFL, the peak von Mises stress in the fifth metatarsal was 20.74 MPa, while with laxity in the CFL, it was 17.52 MPa. However, when both ligaments were relaxed simultaneously, the peak von Mises stress surged to 21.93 MPa. When the ATFL exhibits laxity, the CFL is subjected to a higher stress of 3.84 MPa. Conversely, when the CFL displays laxity, the ATFL experiences a peak von Mises stress of 2.34 MPa.

Conclusion: This study found that changes in the laxity of the ATFL and the CFL are linked to shifts in metatarsal stress levels, potentially affecting ankle joint stability. These alterations may contribute to the progression towards CLAI in individuals with posterolateral ankle ligament injuries. Additionally, significant muscle activation pattern changes were observed in CLAI patients, suggesting altered neural control strategies post-ankle ligament injury.

KEYWORDS

lateral ankle ligament injury, finite element analysis, foot, ligament mechanics, composite material

1 Introduction

Lateral Ankle Ligament (LAL) injuries are among the most prevalent injuries affecting the ankle joint, with occurrences noted across various populations and levels of athletic activity (Swanson et al., 2022). According to research statistics, LAL injuries represent approximately 85% or more of all ankle injuries (Ferran and Maffulli, 2006). Among these, a prevalent type of LAL injury is the anterior talofibular ligament (ATFL) injury, often occurring concomitantly with injuries to the calcaneofibular ligament (CFL) or both the calcaneofibular ligament and posterior talofibular ligament (PTFL) (DiGiovanni et al., 2004; Maffulli et al., 2008). Injuries to the ATFL represent 85% of these injuries, while injuries to the CFL account for 35%. Extending from the anterior margin of the lateral fibular ankle to the neck of the talus, ATFL is also recognized as the structurally weakest among the lateral ankle ligaments (Khawaji and Soames, 2015). Despite a 60%–80% healing rate with conservative treatment for ankle sprains, cases where injuries fail to heal properly may lead to chronic lateral ankle instability (CLAI) (Hintermann et al., 2002). In this case, altered kinematics of the tibial talonavicular joint and increased cartilage contact stresses lead to degenerative joint disease (Bischof et al., 2010).

Injury to the LAL not only induces ligament structural alterations but may also exert diverse impacts on the neuromuscular system. Research indicates that following LAL injury, patients may experience diminished neuromuscular recruitment, weakened muscle strength, and impaired motor coordination (Gribble et al., 2013). These consequences can manifest as challenges in daily activities, including uncoordinated gait patterns during walking and running, or even difficulties in maintaining postural stability. Damage to the neuromuscular system not only impacts the patient's quality of life but also directly influences their motor performance and athletic capabilities (Xu et al., 2023a). Diminished neuromuscular recruitment is a common consequence following LAL injury (Doherty et al., 2014). Ligament injuries can result in reduced joint proprioception, affecting the nervous system's perception and responsiveness, consequently diminishing the sensitivity and responsiveness of muscles to nerve signals (Kunugi et al., 2017). This can decrease muscle activation, disrupting normal motor control and movement coordination. Moreover, LAL injuries may contribute to a decline in muscle strength. Because ligament injuries induce joint instability, the adjacent muscles may undergo functional deterioration, resulting in diminished muscle strength (Lin et al., 2019; Han et al., 2022). This reduction in muscle strength can exacerbate joint instability, perpetuating a cycle of vulnerability to injury (Xu et al., 2023b).

Studies have shown that the LAL plays a crucial role in maintaining ankle stability, and injury to it may lead to impaired joint stability (Gribble et al., 2016). Patients may experience ankle laxity and lateral instability following injury, increasing the risk of re-sprain or recurrence (Xu et al., 2024). Normally, the LAL maintains stability by supporting the ankle and limiting joint motion. However, once the LAL is damaged, the joint's stability will be compromised. Secondly, the loss of ankle stability not only affects the patient's daily activities but may also increase the risk of joint damage (Hintermann et al., 2004). Due to the lack of joint stability, patients are more susceptible to external impact or torque during sports or activities, increasing the likelihood of re-injury (Bae et al., 2015). Ankle instability may also damage the periarticular soft tissues, exacerbating joint dysfunction and pain symptoms. Chronic ankle instability can ultimately result in cartilage degeneration and the eventual development of ankle osteoarthritis (Valderrabano et al., 2006). Long-term follow-up studies have reported that osteoarthritis occurs in 13%–78% of patients with ankle instability over 10 years (Valderrabano et al., 2009). Additionally, recent studies involving patients have identified LAL injuries as the primary cause of ankle osteoarthritis following ligamentous lesions (Löfvenberg et al., 1994).

The finite element (FE) method plays an important role in the field of biomechanics, enabling in-depth analysis of the behavior of complex joints and tissues under clinically relevant loading conditions. It not only accurately simulates real-life scenarios but also provides insights beyond traditional biomechanical studies. When analyzing ankle stability, finite element modeling emerges as a powerful tool for studying the mechanical behavior of the lateral ligament of the ankle. The material behavior of both the ankle and the lateral collateral ligament typically exhibits nonlinearity and nonuniformity, particularly under conditions involving significant deformations and high strain rates (Telfer et al., 2014). Finite element models can integrate these material properties, thereby enabling a more precise simulation of the stress-strain response of the ligaments, as well as the deformation behavior under varying loading conditions (Peng et al., 2023). By meticulously modeling the geometry and material properties of the ankle joint, finite element simulations offer researchers valuable insights into the mechanisms governing ankle stability. Additionally, they unveil the internal stress distribution and deformation patterns of the ligaments under stress conditions (Mabrouk et al., 2022).

There have been previous finite element studies on the ankle joint. Mangwani et al. utilized finite elements to analyze the impacts of ankle ligament injuries, including lateral, syndesmotic, and medial injuries (Talbott et al., 2023). They also investigated the effects of stepwise repairs for each injury on joint displacements and contact stresses, providing valuable insights into optimal repair strategies and prognosis (Halloran et al., 2023). Furthermore, finite element modeling can optimize techniques such as determining the number and placement of bone anchors in lateral ligament reconstruction, evaluating the impact of fibrous bands or hamstring reinforcement on lateral ligament reconstruction, and dynamically stabilizing screws in the joint (Shin et al., 2012). It can also quantify the effects of bone alignment, such as heel pronation and correction. Additionally, finite

TABLE 1 Participant demographics.

	CLAI (n = 20)	Healthy (n = 20)	P
Age (year)	23.8 ± 1.5	24.1 ± 1.3	0.441
Mass (kg)	82.3 ± 4.9	81.1 ± 6.1	0.738
Height (cm)	181.3 ± 3.2	179.6 ± 5.4	0.535
Ankle sprains (times)	3.4 ± 1.2	0	<0.001
Time since injury (month)	10.2 ± 2.3	0	<0.001
Leg length (cm)	95.7 ± 4.5	92.9 ± 5.2	0.769

CLAI, chronic lateral ankle instability.

element analysis can serve as a treatment and reattachment strategy for insertional Achilles tendinopathy (Tits and Ruffoni, 2021).

Previous research has not explored the changes in joint stability and stress response distribution in patients with ankle ligament injuries during landing, understanding these changes is crucial for developing effective rehabilitation strategies and preventive measures for ankle ligament injuries. Therefore, this study aimed to explore the stress response and neural control changes during landing in individuals with lateral ankle ligament injuries and to evaluate the impact of varying degrees of ligament laxity on metatarsal stress. Our hypothesis posited that metatarsal stress would escalate with increased ligament laxity.

2 Method

The main framework of this study consists of the following components: 1) Collection of subject biomechanical data. 2) Processing data using Matlab. 3) Construction of finite element models. 4) Validation of finite element model results using a high-speed dual fluoroscopic imaging system (DFIS). First, kinematic and kinetic data of the participants were collected using Kistler force plates and Vicon, while sEMG signals were gathered using EMG sensors. Next, the data were preprocessed using MATLAB. Then, a finite element model was constructed, and the data were imported. Finally, the results of the finite element model were validated using DFIS.

3 Subjects

Twenty healthy subjects and twenty subjects with CLAI were recruited for this study (Table 1). All CLAI subjects were treated conservatively for 6 months before the experiment; however, they continued to exhibit symptoms of pain, instability and decreased proprioception. The diagnosis of CLAI was confirmed through a clinical examination by a foot and ankle orthopedic surgeon, and the clinical manifestations of lateral ankle injury were confirmed by magnetic resonance imaging (MRI). The criteria for subject screening were: 1) no history of foot and ankle surgeries; 2) no additional ankle pathology was identified by MRI. Reasons for exclusion included peroneal tendinopathy, and intra-articular small bone or cartilage degeneration (Caputo et al., 2009).

Prior to engaging in the study, all individuals were thoroughly informed on the study's purpose, requirements, and procedures. This information was provided after each participant had signed the informed consent form. The study protocol received approval from the Ningbo University Scientific Research Ethics Committee (Approval Number: RAGH20240405).

3.1 Biomechanics parameters collection and processing

This study utilized a Kistler force platform and an eight-camera Vicon motion capture system (Oxford Metrics Ltd., Oxford, United Kingdom) to synchronize the kinetic and kinematic data collection. Kinematic data collection was performed at frequencies of 100 Hz, while kinetic data collection was conducted at frequencies of 200 Hz (Ward et al., 2009). Markers were positioned according to the gait 2,392 (Figure 1A), and EMG sensors were placed following the guidelines provided by SENIAM (Figure 1C) (Hermens et al., 2000). This study minimized the impedance of the skin-electrode interface by shaving the hair near the skin and cleaning the area with alcohol. Muscle activation was measured using six electromyography (EMG) sensors from Delsys (Boston, MA, United States) (Figure 2). All subjects were instructed to perform landings from two consecutive steps to collect the corresponding kinetic data. Data was successfully collected 20 times for each subject, recording only the data from stable landings (Figure 1B). The three-dimensional marker trajectories and ground reaction force data were collected using Vicon Nexus 2.14.0 and exported as C3D format files. These data were then processed in MATLAB (MathWorks, MA, United States), which involved coordinate system conversion, low-pass filtering, data extraction, and format conversion for both the kinematic and ground reaction force data. This study utilized the OpenSim software (Stanford, CA, United States) for finite element analysis to calculate biomechanical parameters as the boundary condition (Cheung et al., 2009).

3.2 The process of obtaining and reconstructing geometric data

One of the subjects with chronic lateral ankle instability (CLAI) was chosen to undergo magnetic resonance imaging (MRI) and computed tomography (CT) with a 2 mm interval for this study. The resulting 2D images were then segmented using Mimics 21.0 (Materialise, Leuven, Belgium). Subsequently, the generated foot model was imported into Geomagic Studio 2021 (Geomagic, Inc.,

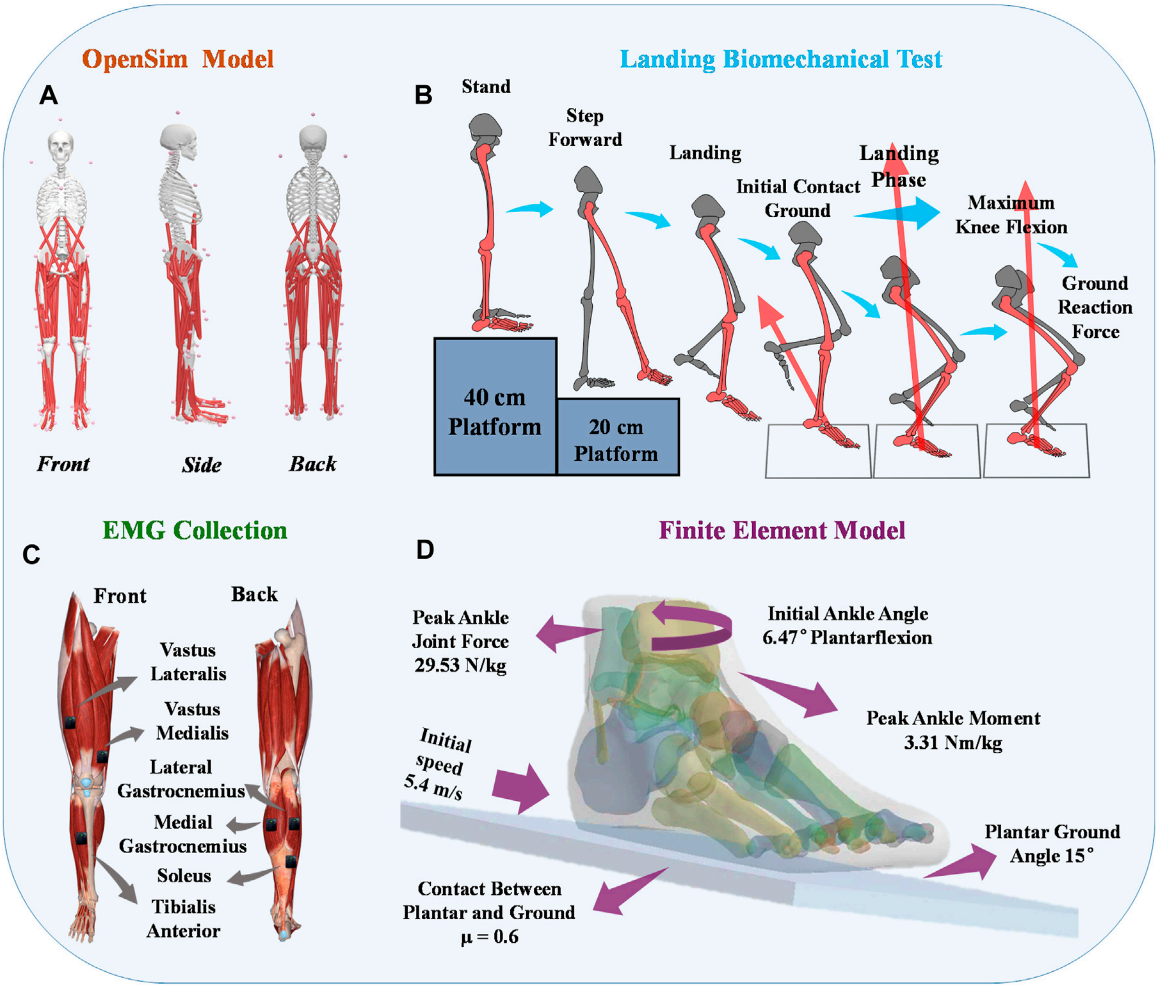

FIGURE 1
Illustrative representation depicting the placement of reflective marker points on the musculoskeletal model. **(A)** Visual representation of the placement of reflective markers on the constructed musculoskeletal model. **(B)** Depiction of the positioning of EMG electrodes on the lower limbs of human subjects. **(C)** Illustration outlining the procedure of the unanticipated landing biomechanics test. **(D)** Finite Element Boundary Condition Setting.

Research Triangle Park, NC, United States) to optimize the model. The imported components were then assembled into solid form using SolidWorks 2017 (SolidWorks Corporation, Waltham, MA, United States). Finally, the model's contacts were meshed and modeled using Workbench 2021 (ANSYS, Inc. located in Canonsburg, PA, United States), and static finite element analysis was conducted (Figure 1D).

3.3 Composite representation of the LAL

In the finite element model, the lateral collateral ligament (LAL) of the ankle was initially simulated too homogeneously, and this simplistic representation could not accurately capture the laxity of the lateral ankle ligament. Setting the lateral ankle ligament to be isotropic may lead to inaccurate results (Siegler et al., 1988). Therefore, we utilized a refined hyperelastic composite to model the physiological structure of the LAL, incorporating collagen fibers for enhanced accuracy (Figure 3A). This composite LAL model comprises a proteoglycan matrix reinforced by collagen fibers, with a collagen fiber volume fraction set at 60% (Kumai et al., 2002). This value was determined based on the upper limit of the total cross-sectional area occupied by collagenous proto-fibers in the LAL.

The strain energy formula for hyperelastic material is derived from the strain energy density function:

$$\Psi = \Psi_{vol}(J) + \Psi_{iso}^{m}(\bar{C}) + \Psi_{iso}^{f}(\lambda) \tag{1}$$

In this study, a Neo-Hookean model was employed. The volumetric component, denoted as Ψ_{iso}, represents the change in ligament volume. Additionally, Ψ_{iso} denotes the deviatoric component of shape change, where Ψ_{iso}^{m} corresponds to the Stromal Fraction and Ψ_{iso}^{f} represents the fiber fraction. J stands for the Jacques ratio of the deformation gradient tensor F, while $\bar{C}$ signifies the deviatoric component of the deformation gradient tensor C ($\bar{C} = J^{-\frac{2}{3}}C$) (Weiss et al., 2002). Moreover, the attachment zones centers at both ends of the ligament served as the initial direction of the fiber. The elongation λ was computed based on the material's deformation and the initial fiber direction a_0 ($\lambda = \bar{C} \times a_0^2$).

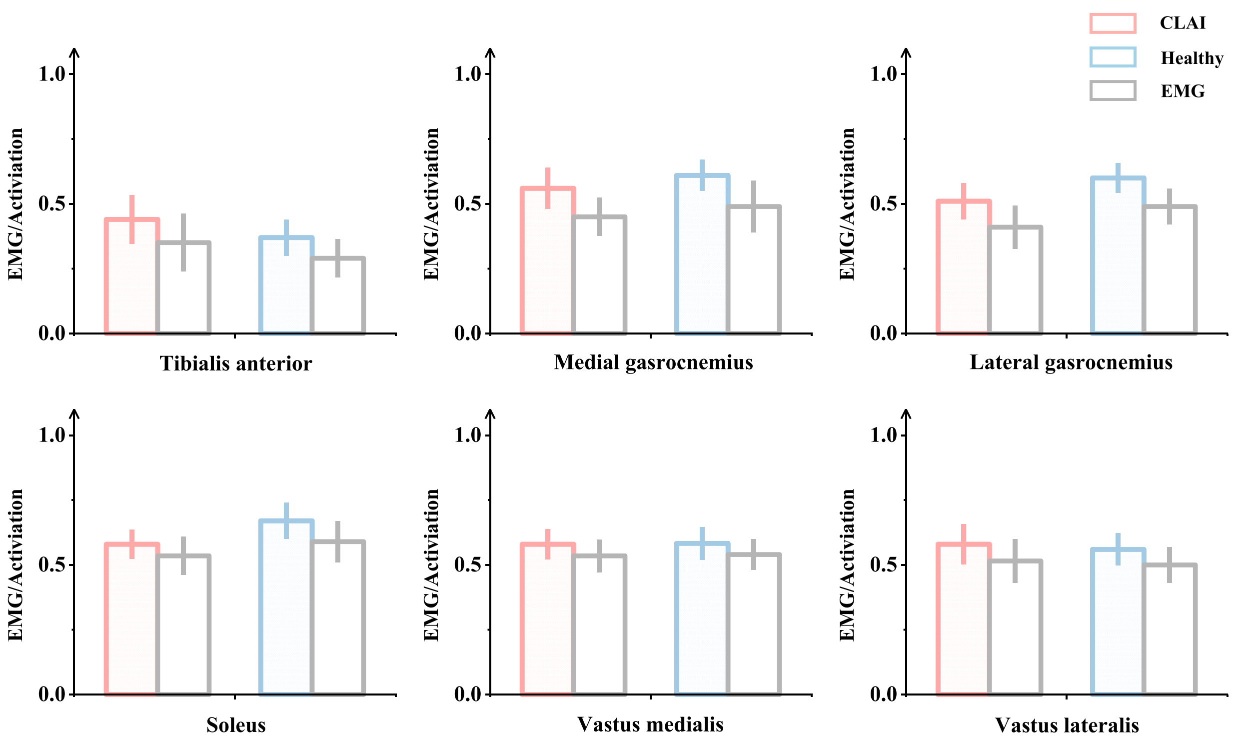

FIGURE 2
Validation was conducted by comparing the patterns of measured muscle activation and simulated activation.

For the volumetric component Ψ_{vol}:

$$\Psi_{vol}(J) = \frac{1}{2D}\ln J^2 \tag{2}$$

For the deviatoric component $\Psi^m_{iso}(\bar{C})$:

$$\Psi^m_{iso}(\bar{C}) = C_1(\bar{I_1} - 3) \tag{3}$$

Limited ability of collagen fibers in the ligament to withstand loads, so it is defined as (Weiss et al., 1996):

$$\Psi^f_{iso}(\lambda) = F_2(\lambda) \tag{4}$$

Therefore, the strain energy equation is determined as:

$$\Psi = \frac{1}{2D}\ln J^2 + C_1(\bar{I_1} - 3) + F_2(\lambda) \tag{5}$$

3.4 Boundary, loading condition and model validation

To simulate the real situation, in this study, the angle of the ankle joint was determined based on the position of the foot model. This was achieved by fixing the floor and adjusting the angle between the tibial axis and the longitudinal axis of the foot in the sagittal plane within the finite element model. The adjustment ensured alignment between the global coordinate system and the original coordinate system of OpenSim (Delp et al., 2007). The inertial forces experienced during landing were simulated by applying ankle moments and reaction forces to the talar sliding joint and tibiotalar joint surfaces, respectively. Additionally, joint forces of the MPJ were applied to the upper surface of the first metatarsal and proximal phalanx bone. All materials were assumed to be isotropic and linearly elastic, except for the encapsulated soft tissues, skin, and ligaments, for which properties were derived from previous studies (Table 2). In this study, fluoroscopic image data of the ankle joints of subjects in the grounded condition were acquired using a high-speed dual fluoroscopic imaging system (DFIS) (Ti-WISH-II, Ti-Motion Ltd., Shanghai City, CN). These computed ankle joint displacements were then compared with the results obtained from finite-element analysis to validate its accuracy (Roach et al., 2017). Specifically, we validated the accuracy of the finite element model of the foot by measuring the displacement of the navicular bone, which is commonly used in clinical practice to assess specific deformations of the foot.

First, high-resolution images of the ankle joint were obtained using DFIS imaging technology to ensure clarity and accuracy. These images were then imported into Rhinoceros software for detailed model adjustments. Next, the displacement of the navicular bone was calculated using the coordinate system calculator plugin in Rhinoceros (Li et al., 2008). This plugin accurately measures the spatial position changes of various points in the model to compute the actual displacement data of the navicular bone. Finally, these calculated results were compared with the finite element model for verification. The finite element model provides theoretical displacement data, and by comparing it with the actual calculated displacement data, the accuracy of the model can be validated (Figure 3B) (Nielsen et al., 2009; Koo and Li, 2016).

DFIS system generates high voltage through a high-voltage generator, and the X-ray tube is used to generate X-rays. The X-ray tube contains a cathode and an anode, when high voltage

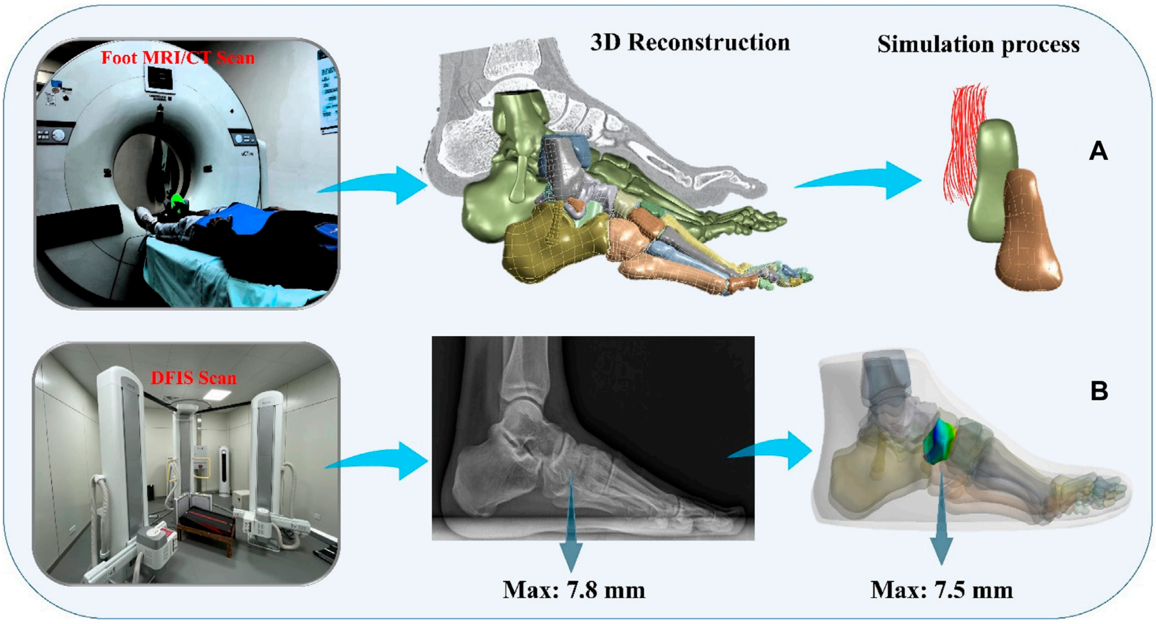

FIGURE 3
Finite element simulation workflows. **(A)** Composite representation of the anterior talofibular ligament; **(B)** Finite element model validation.

TABLE 2 Material properties of the components in the finite element model.

Component	Element type	Elastic modules (MPa): *E*	Poisson's ratio: *v*	Destiny(kg/m^3)	Reference
Skin	Tetrahedral solid	Hyperelastic (first-order Ogden model, $\mu = 0.122\,kPa, \alpha = 18$)	N/A	950	Pailler-Mattei et al. (2008)
Bulk soft tissue	Tetrahedral solid	Hyperelastic (second-order polynomial strain, $C_{10} = 0.8556, C_{01} = 0.05841, C_{20} = 0.03900, C_{11} = 0.02319, C_{02} = 0.00851, D_1 = 3.65273$)	N/A	950	Gu et al. (2010)
Bones	Tetrahedral solid	7,300	0.3	1,500	Cheung et al. (2005)
Cartilages	Tetrahedral solid	1	0.4	1,050	
Ligaments	Two-node truss	260	0.4	1,000	Kitaoka et al. (1994)
ATFL PTFL CFL	Tetrahedral solid	Hyperelastic (second-order polynomial strain, $C_{10} = -222.1, C_{01} = 290.97, C_{20} = -1.1257, C_{11} = 4.7267, C_{02} = 79.602$	N/A	1,000	Peng et al. (2023)
Achilles Tendon	Two-node truss	816	0.3	1,000	Chen et al. (2012)
Plate	Hexahedral solid	17,000	0.4	1,000	Xiang et al. (2022)

ATFL, anterior talofibular ligament; PTFL, posterior talofibular ligament; CFL, calcaneofibular ligament.

is applied to the X-ray tube, the cathode releases free electrons. These changes produce an image on the detector, forming an X-ray fluoroscopic image. The DFIS system is equipped with two X-ray sources and two detectors, which are positioned on each side of the object under observation. This setup allows for the simultaneous acquisition of X-ray fluoroscopic images from both directions, enabling reliable medical image data for clinical diagnosis.

4 Result

This study investigated the kinematic and kinetic changes in healthy individuals and patients with CLAI during landing. Additionally, we explored the stress distribution using a finite element model. The specific findings are detailed as follows:

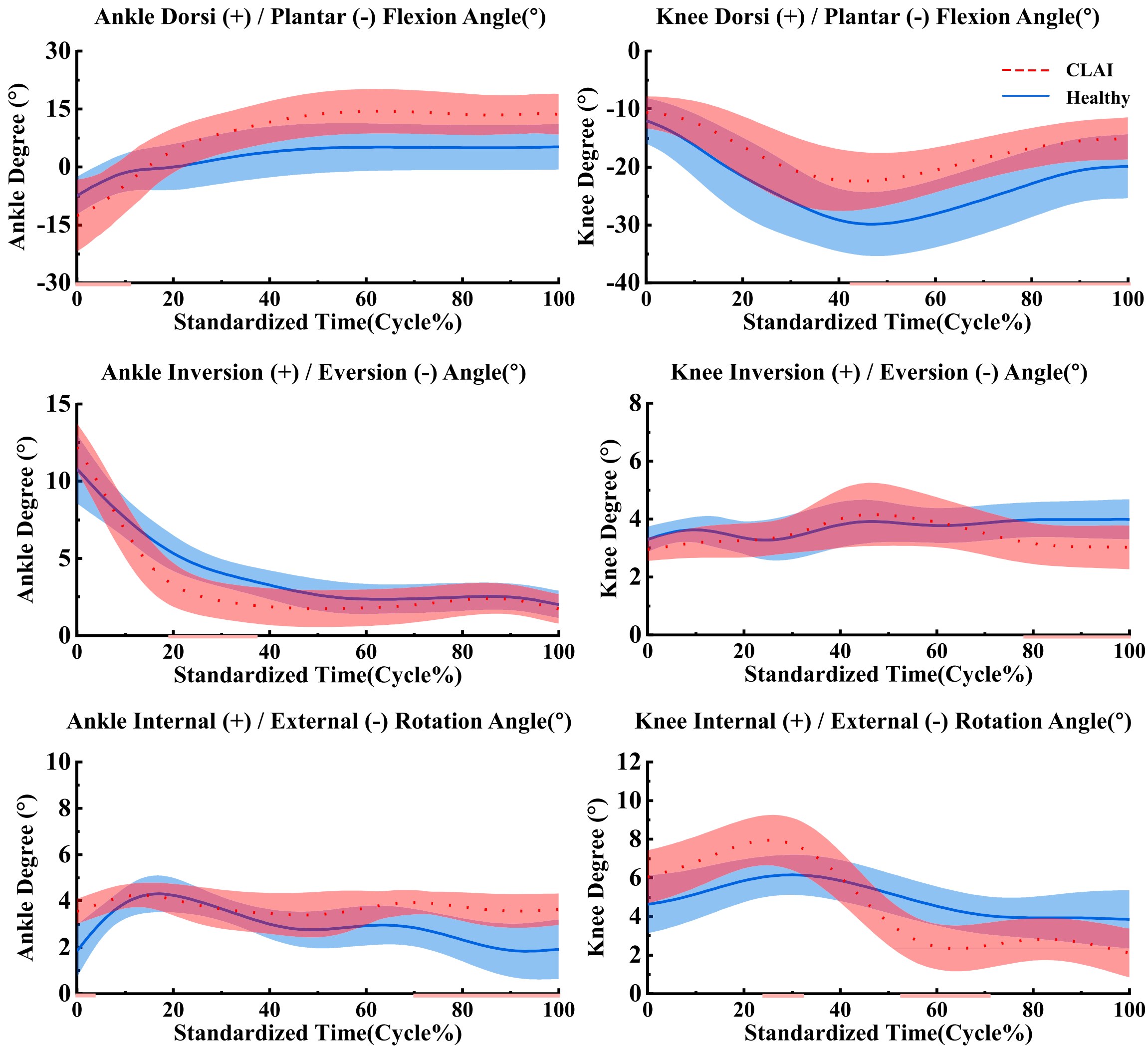

FIGURE 4
Mean and standard deviation of the ankle and knee joints. The red line illustrates the SPM analysis findings between the CLAI and healthy groups. From top to bottom are sagittal plane, coronal plane and horizontal plane.

4.1 Joint angle and moment

According to Figure 4, during the 0%–14% landing phase, the ankle joint angle in the sagittal plane shows significant changes in CLAI patients compared to the healthy group ($p < 0.001$). In the coronal plane, during the 19%–38% landing phase, the ankle joint angle changes in the healthy group are significantly greater than those in CLAI patients ($p = 0.033$). In the horizontal plane, during the initial 0%–3% landing phase, the ankle joint angle changes in CLAI patients are significantly greater than in the healthy group ($p = 0.016$). Additionally, during the 70%–100% landing phase, the ankle joint angle in CLAI patients shows significant changes compared to the healthy group ($p < 0.001$).

For the knee joint, in the sagittal plane, CLAI patients are greater than that of the healthy group. Significant changes in the knee joint angle of CLAI patients are observed during the 42%–100% landing phase ($p < 0.001$). In the coronal plane, during the 77%–100% landing phase, the knee joint angle changes in the healthy group are significantly greater than those in CLAI patients ($p < 0.05$). In the horizontal plane, during the 23%–31% landing phase, the knee joint angle changes in CLAI patients are significantly greater than in the healthy group ($p < 0.001$), while during the 52%–71% phase, the angle changes are significantly smaller than in the healthy group ($p < 0.05$) (Table 3).

The results indicate that in the sagittal plane, the peak ankle joint moment in the CLAI group is significantly smaller than in the healthy group ($p < 0.01$). In contrast, in the coronal plane, the peak ankle joint moment in the CLAI group is significantly higher than in the healthy group ($p < 0.05$). However, in the horizon plane, there was no significant difference between the two groups ($p = 0.05$). For the knee joint moment in the sagittal plane, the CLAI group showed significantly higher compared to the healthy group ($p < 0.001$). Similarly, in the coronal plane, the knee joint moment was significantly greater in the CLAI group than in the healthy group ($p < 0.01$). Differences were also observed in the horizon plane, with the CLAI group exhibiting higher knee joint moments than the healthy group ($p < 0.05$) (Table 3).

TABLE 3 Detailed biomechanical results between CLAI and Healthy group.

Parameters	CLAI	Healthy	P	ES
Sagittal plane peak ankle moment (Nm)	0.93	1.08	<0.01	0.274
Coronal plane peak ankle moment (Nm)	0.57	0.48	<0.05	0.162
Horizon plane peak ankle moment (Nm)	0.44	0.39	0.05	0.121
Sagittal plane peak knee moment (Nm)	1.87	1.69	<0.001	0.398
Coronal plane peak knee moment (Nm)	0.83	0.66	<0.001	0.285
Horizon plane peak knee moment (Nm)	0.61	0.55	<0.05	0.212
Sagittal plane peak ankle angle (°)	12.87	6.51	<0.001	0.863
Coronal plane peak ankle angle (°)	12.44	10.37	<0.05	0.195
Horizon plane peak ankle angle (°)	4.12	4.19	0.68	0.012
Sagittal plane peak knee angle (°)	−21.78	−29.95	<0.001	0.531
Coronal plane peak knee angle (°)	3.86	3.72	0.05	0.117
Horizon plane peak knee angle (°)	8.03	5.74	<0.001	0.416

ES, effect size; CLAI, chronic lateral ankle instability.

4.2 Muscle activation

Patients with chronic lateral ankle instability (CLAI) exhibit different patterns of muscle activation compared to the healthy group during landing. Based on Figure 5, the muscle activation of the vastus lateralis muscle during landing was higher in CLAI patients compared to the healthy group ($p < 0.05$). Additionally, the results revealed that during the initial 0%–12% phase of landing, the muscle activation of the vastus medialis muscle in CLAI patients preceded that of the healthy group ($p < 0.001$), with higher peak muscle activation observed as well. However, injury to the lateral ankle ligament during landing resulted in significant differences. Conversely, during the 8%–35% landing phase ($p = 0.012$), the healthy group exhibited significantly higher muscle activation in the soleus compared to the CLAI group. Regarding the medial gastrocnemius and lateral gastrocnemius muscles, the results demonstrated a similar disparity in muscle activation between the healthy group and the CLAI group. Muscle activation in the medial gastrocnemius exhibited an earlier peak activation in the healthy group ($p < 0.001$), along with a higher peak muscle activation. Similarly, activation of the lateral gastrocnemius muscle occurred earlier in the healthy group compared to the CLAI group ($p < 0.001$). Furthermore, significant differences were observed in the tibialis anterior muscle between the CLAI and healthy groups during the 22%–40% and 48%–61% landing phases ($p < 0.05$).

4.3 Muscle force

Based on SPM1d analysis, according to Figure 6, significant differences were observed in the muscle force of the tibialis anterior muscle between the healthy group and CLAI patients during the 0%–36% landing phase ($p < 0.05$). Conversely, during the 81%–98% phase of landing, the CLAI group exhibited higher muscle force compared to the healthy group. For the soleus, the CLAI group was higher than the healthy group most of the time, notably demonstrating a significant difference during the 30%–40% phase ($p < 0.05$). However, it was lower than the healthy group during the 0%–19% and 93%–100% landing phases ($p < 0.001$, $p < 0.05$, respectively). The muscle force of the peroneus longus muscle in the CLAI group was comparable to that of the healthy group, except for the 32%–44% phase of the landing, where it was lower than the healthy group ($p < 0.001$). Meanwhile, the medial gastrocnemius muscle force was significantly higher in the healthy group than in the CLAI group during the 0%–16% landing phase ($p < 0.001$), while in the 42%–78% phase ($p < 0.001$) it was higher in the CLAI group than in the healthy group. Lateral gastrocnemius muscle force was essentially similar between the two groups, although differences were observed during the 0%–5% and 21%–32% landing phases ($p = 0.034$, $p < 0.05$, respectively).

4.4 Stress distribution results

During the simulation of a CLAI patient's landing, the thermograms generated from finite element analysis illustrate changes in stress distribution on the ligaments as different ligaments exhibit laxity (Figure 7). When the ATFL exhibits laxity, the peak von Mises stress in the CFL rises to 3.8359 MPa, while the peak von Mises stress within the ATFL itself is 1.7911 MPa. A similar situation was observed when the CFL exhibited laxity. The stress on the ATFL increased significantly to 2.3381 MPa compared to its laxity. The CFL did not experience any additional stress and only 2.1875 MPa. However, some differences emerge when the ATFL and CFL exhibit laxity. The stresses experienced by the ATFL and CFL do not increase, with the peak von Mises stress on the ATFL being 1.3351 MPa, and on the CFL being 1.9835 MPa. Figure 8 illustrates the effect of ligament laxity on the calcaneus and talus bones. Under conditions where the

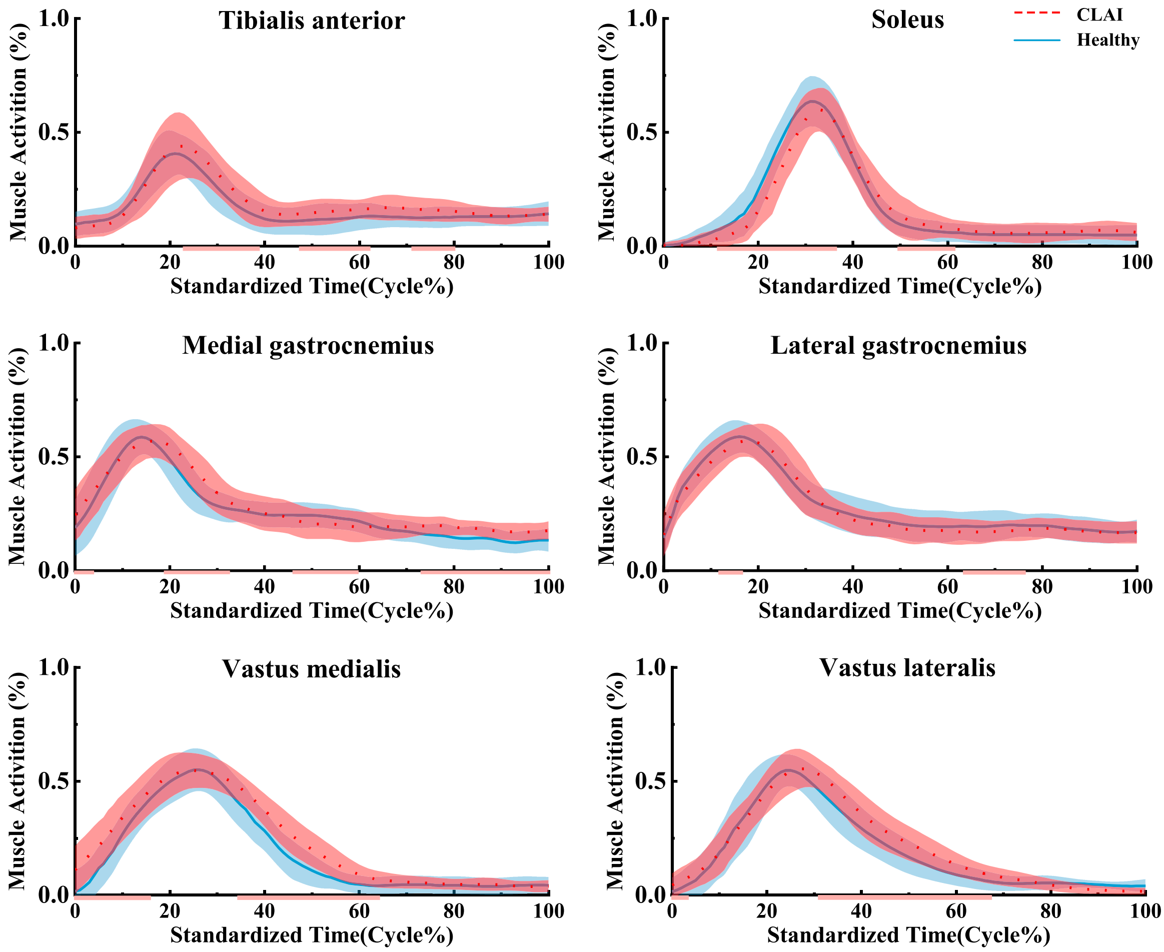

FIGURE 5
Mean and standard deviation of the simulated muscle activations. The red line illustrates the SPM analysis findings between the CLAI and healthy groups.

CFL exhibits laxity, the peak von Mises stress on the calcaneus is 7.6799 MPa. However, when the ATFL exhibits laxity, the stress on the calcaneus increases to 8.5861 MPa, and the overall stress distribution is higher than when the CFL exhibits laxity. When both the ATFL and CFL exhibit laxity, the peak von Mises stress on the calcaneus significantly increases to 9.1576 MPa. The stress distribution of the talus bone shows a similar trend. When the ATFL exhibits laxity, the peak von Mises stress on the talus bone is 4.8085 MPa, while under CFL laxity, the stress on the talus bone is slightly lower at 3.6159 MPa. When both ligaments exhibit laxity, the peak von Mises stress on the talus bone increases to 5.2019 MPa.

Figure 9 illustrates the changes in stress on different metatarsal bones when ligaments exhibit laxity. The stress variation is relatively small in the first metatarsal bone, with a peak von Mises stress of 11.9532 MPa observed when the ATFL exhibits laxity. Under CFL laxity conditions, the peak von Mises stress measures only 10.1714 MPa, slightly lower than observed with ATFL laxity. However, with simultaneous laxity in both ligaments, the peak von Mises stress reaches 13.898 MPa. However, the second metatarsal bone exhibits a different pattern. With the ATFL laxity, the peak von Mises stress measures 15.366 MPa, which increases to 17.936 MPa under CFL laxity. When laxity occurs in both ligaments simultaneously, the peak von Mises stress rises to 20.201 MPa. The third metatarsal bone demonstrates a behavior that resembles that of the second metatarsal bone, showing a peak von Mises stress of 10.183 MPa under ATFL laxity and 12.537 MPa under CFL laxity. With laxity present in both ligaments, the peak von Mises stress increases to 16.673 MPa. The variation in peak von Mises stress is less noticeable in the fourth metatarsal bone, with 13.548 MPa when the ATFL exhibits laxity, 10.506 MPa when the CFL exhibits laxity, and 15.096 MPa when both ligaments exhibit laxity. The fifth metatarsal bone exhibits the highest von Mises stress among the metatarsals. Under ATFL laxity, the peak von Mises stress measures 20.740 MPa, while under CFL laxity, it measures 17.521 MPa. Simultaneous laxity in both ligaments results in a peak von Mises stress of 21.931 MPa.

5 Discussion

The purpose of this study was to examine alterations in stress response and neural control during landing among patients with lateral collateral ligament injuries of the ankle. We conducted a comparative analysis of metatarsal stress changes during landing in

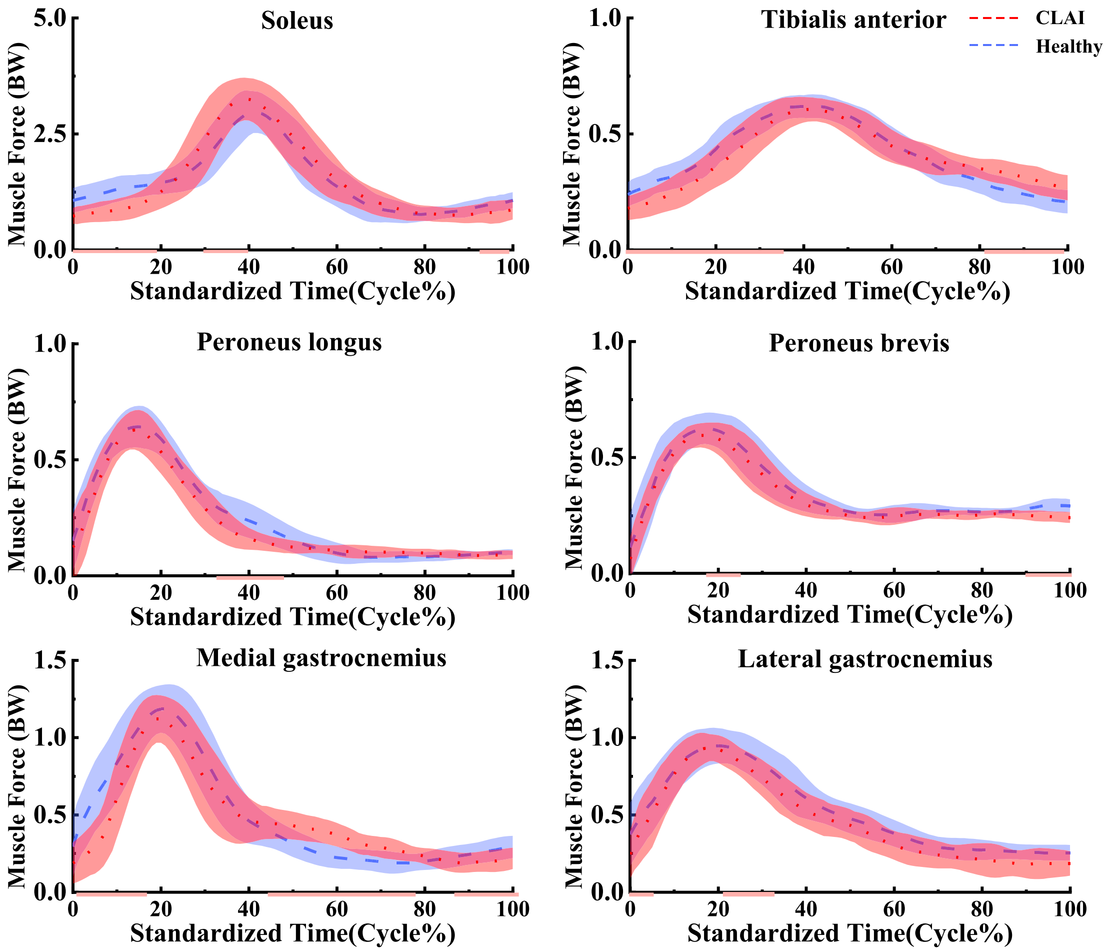

FIGURE 6
Mean and standard deviation of the simulated muscle forces. The red line illustrates the SPM analysis findings between the CLAI and healthy groups.

subjects exhibiting varying degrees of lateral collateral ligament laxity in the ankle, aiming to enhance ankle stability in patients. Additionally, we aimed to contribute to the development of more scientifically informed treatment protocols for the rehabilitation of such patients through further research. We hypothesized that metatarsal stress would vary with the degree of laxity of the lateral ligament of the ankle joint, thus affecting the stability of the ankle joint. The results of this study are consistent with our initial hypothesis.

The study found that compared to the healthy group, CLAI patients exhibited greater ankle joint angle changes in the sagittal plane. This difference may be due to their inability to effectively regulate ankle joint angles upon landing, leading to abnormal joint movement patterns. Additionally, there were differences in knee joint angles between the healthy group and CLAI patients, indicating distinct knee joint movement control strategies between the two groups. These findings are consistent with previous research, suggesting that CLAI patients display altered landing patterns compared to the healthy group (Brown et al., 2008). Gehring et al. (2014) research highlights significant deviations in knee joint angles during landing and cutting movements in the CLAI group compared to the healthy group, emphasizing the importance of considering unanticipated tasks when assessing actual LAL. The study revealed that the CLAI group employed different movement strategies during landing tasks than the healthy group. Further research is needed to verify the effectiveness of the observed changes in neural control strategies in reducing the risk of LAL.

This study observed that laxity of the lateral ankle ligaments increases stress on the metatarsals, leading to ankle instability. Research indicates that damage or laxity of the lateral ligaments alters the biomechanical properties of the ankle joint. Typically, the lateral ankle ligaments maintain stability by supporting and restricting joint movement (Bonnel et al., 2010; Jiang et al., 2024). Damage to these ligaments may result in increased pressure on the metatarsals, impairing their ability to effectively support ankle movement. The findings also corroborate previous research, showing that further laxity of the ATFL and CFL significantly alters metatarsal stress during simulated CLAI landings. Under CFL laxity, peak von Mises stress on the fifth metatarsal increases, and under ATFL laxity, the peak von Mises stress on the fifth metatarsal is even greater, reaching its peak when both ligaments are lax. Studies by Doherty indicate that lateral ankle ligament injuries restrict inversion and eversion movements of the ankle joint, leading to additional load on the metatarsals and increasing the risk of ankle injury (Doherty et al., 2016). Dobbe et al. observed increased metatarsal stress in patients with chronic

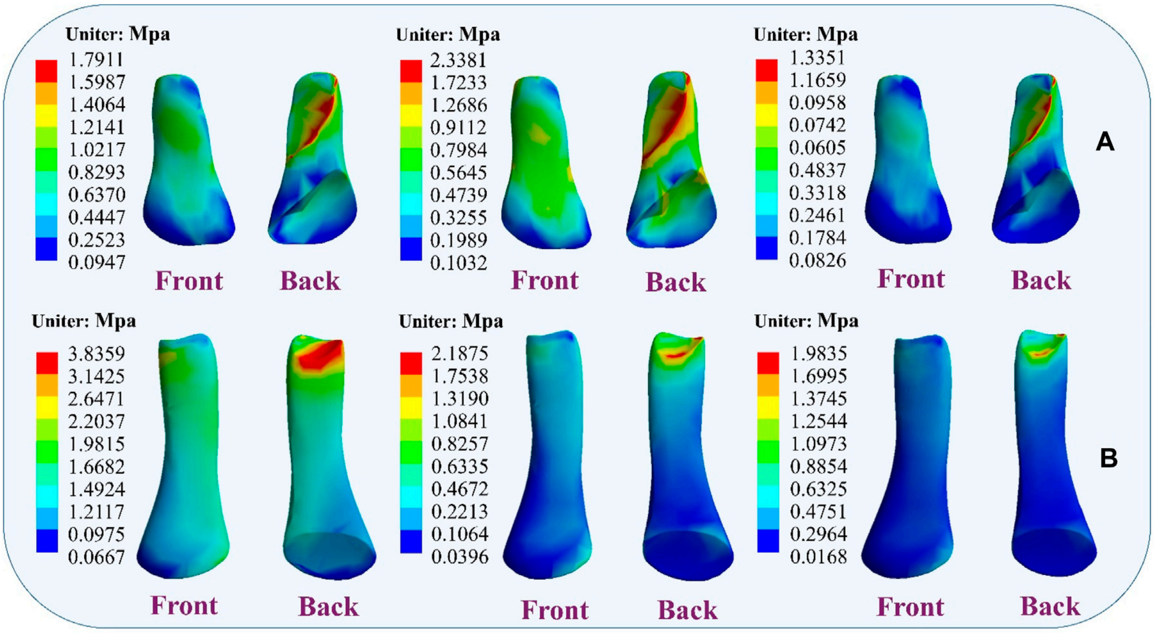

FIGURE 7
Distribution of ligament von Mises stress response during ligament laxity. The results from left to right are anterior talofibular ligament laxity, calcaneofibular ligament laxity, and both ligaments laxity. **(A)** anterior talofibular ligament; **(B)** calcaneofibular ligament.

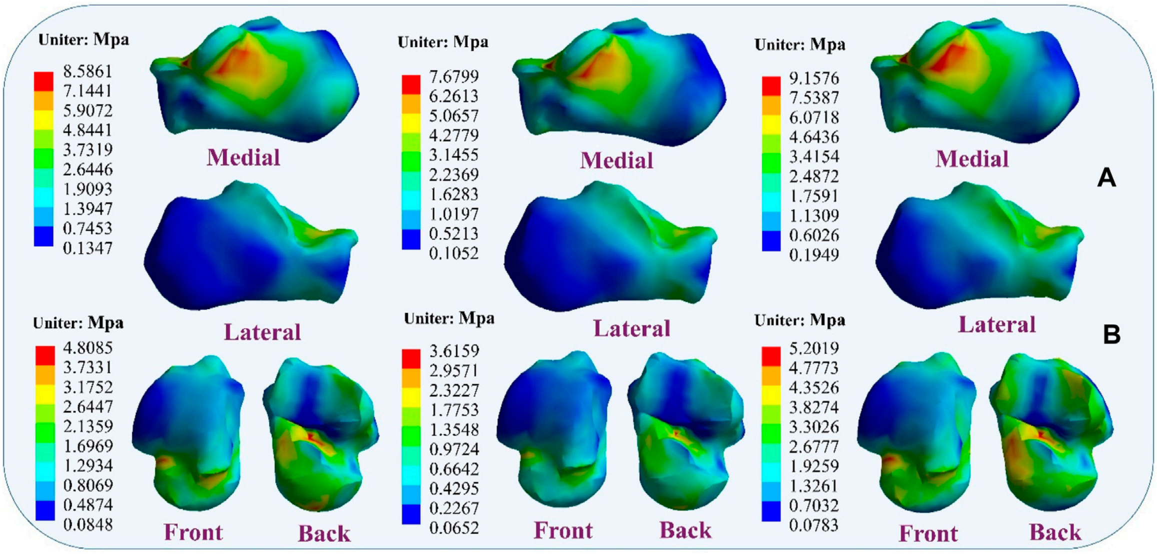

FIGURE 8
Distribution of von Mises stress response during ligament laxity. The results from left to right are anterior talofibular ligament laxity, calcaneofibular ligament laxity, and both ligaments laxity. **(A)** calcaneus bone; **(B)** talus bone.

lateral ankle instability, particularly during physical activity (Dobbe et al., 2020). Additionally, the study results demonstrate that ligament stress gradually increases upon landing with lateral ligament laxity, suggesting changes in the surrounding tissues and structures, consistent with previous findings. During this process, the mechanical properties of the ligaments may change, making them more sensitive to external stress and potentially increasing the risk of secondary injury when damaged ligaments are subjected to external stress (Barelds et al., 2018).

Furthermore, the study found that ligament laxity can induce muscle compensation. Results suggest that laxity of the ATFL may transfer stress to the CFL, and conversely, laxity of the CFL may increase stress on the ATFL. When two ligaments are lax, stress may transfer to other muscles to maintain the stability of the foot

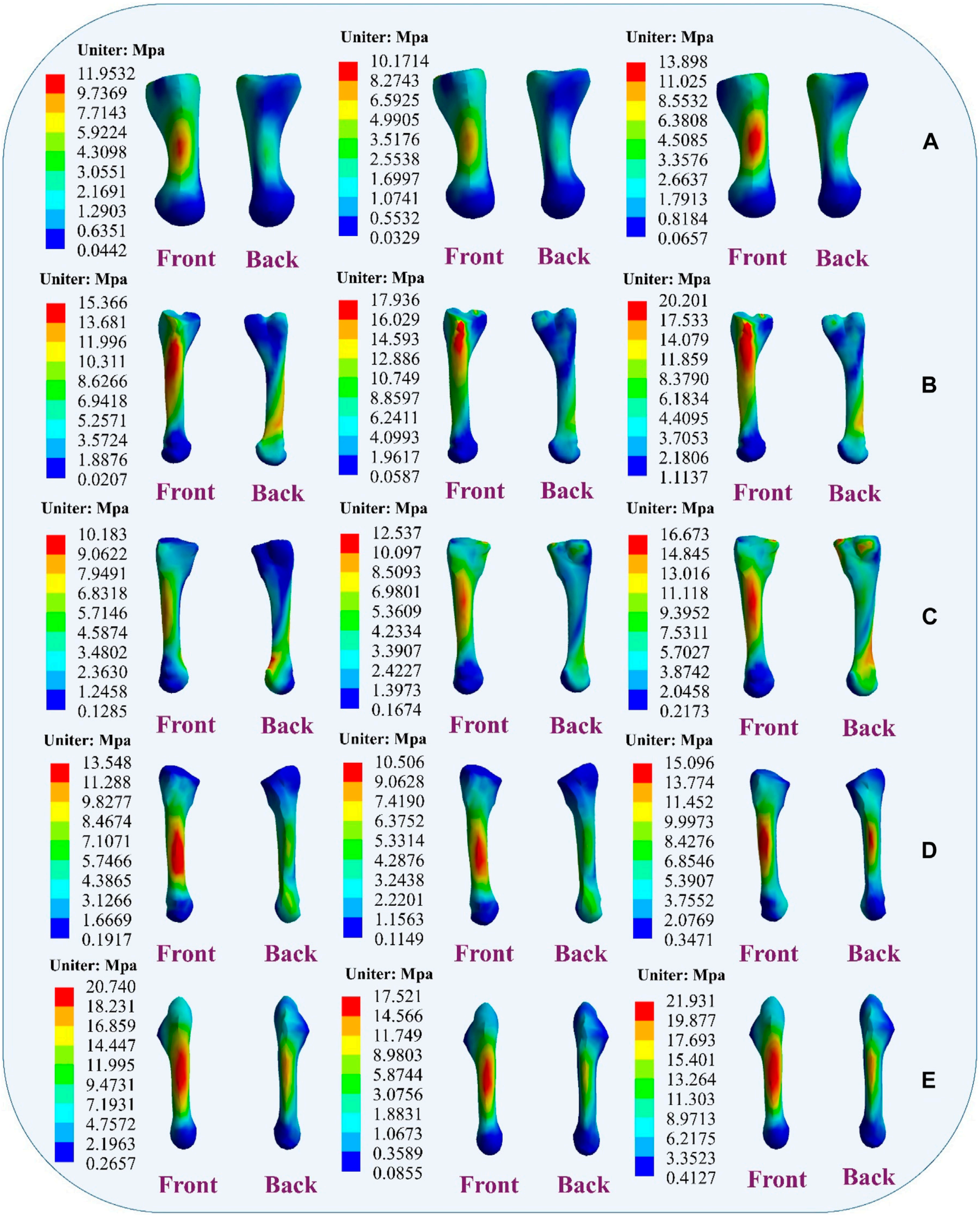

FIGURE 9
Distribution of metatarsal von Mises stress response during ligament laxity. The results from left to right are anterior talofibular ligament laxity, calcaneofibular ligament laxity and both ligaments laxity. **(A)** first metatarsal; **(B)** second metatarsal; **(C)** third metatarsal; **(D)** fourth metatarsal; **(E)** fifth metatarsal.

structure. This muscle compensation may involve muscles and ligaments surrounding the foot. These findings are significant for understanding muscle strength reconstruction during ligament injury rehabilitation and preventing further injury. Banerjee and Agarwal. (1998) research emphasizes the indispensable role of the talus and calcaneus in weight-bearing, providing anchorage for other crucial ligament structures, thereby tightly connecting the critical joints of the distal lower limb. Laxity of the lateral ankle ligaments may further damage the talus and calcaneus. Moreover, the study also found that with ligament laxity, stress between the calcaneus and talus changes, potentially leading to talus displacement and affecting joint stability. Therefore, maintaining the stability of the lateral ankle ligaments is crucial for protecting the talus and calcaneus and preserving the function and stability of the foot joints (Ferran et al., 2009; Kang and Jiang, 2024).

Another noteworthy discovery in the present study is the distinct muscle activation pattern observed in patients with chronic lateral ankle instability (CLAI) during landing compared to the healthy group, which corroborates earlier research. DeJong et al. (2020) demonstrated that individuals with CLAI showcased compensatory muscle activation in the proximal joint, while Rios et al. (2015) observed heightened muscle activation in CLAI patients during single-leg stance compared to healthy controls, particularly in the activation of proximal joint muscles during the standing phase. The current study noted increased muscle activation in the medial and lateral femoral muscles among CLAI patients compared to the healthy group. This indicates that individuals with CLAI tend to rely on heightened proximal muscle activation as a compensatory mechanism to improve motor control and mitigate neuromuscular deficits in the ankle joint. Previous research has similarly highlighted decreased neuromuscular recruitment as a common manifestation following LAL injury (Feger et al., 2014). The ligament injury may lead to decreased joint positional sensation, thereby reducing muscle sensitivity and responsiveness to nerve signals. The study findings indicated delayed activation of the medial gastrocnemius and lateral gastrocnemius muscles in the CLAI group compared to the healthy group. Furthermore, the soleus muscle exhibited not only earlier activation but also higher peak muscle activation. H. Kim's analysis revealed that muscles like the peroneus longus, medial gastrocnemius, and lateral gastrocnemius were activated earlier during walking in the healthy group. Conversely, CLAI patients displayed delayed activation of the lateral gastrocnemius and soleus muscles during landing, potentially contributing to ankle instability (Kim et al., 2019). Moreover, CLAI patients exhibited greater tibialis anterior muscle strength and activation compared to the healthy group, indicating an augmented reliance on ankle dorsiflexion for stabilization during landing (Jie et al., 2024). The above study suggests that CLAI patients exhibit altered motor control patterns during landing compared to the healthy group. In the absence of adequate treatment and rehabilitation, CLAI patients may employ alternative strategies to compensate for ankle instability, potentially heightening the risk of secondary injury (Son et al., 2019).

Given the observed compensatory muscle activation patterns in patients with CLAI, targeted rehabilitation programs can be developed to address these neuromuscular deficits. For instance, strengthening exercises focused on the medial and lateral femoral muscles may help improve proximal muscle support and overall joint stability (Xu et al., 2022; Zhou and Ugbolue, 2024). Additionally, proprioceptive training can be incorporated to enhance joint positional sense and muscle responsiveness, thereby reducing the risk of further injury. By customizing rehabilitation protocols to address specific muscle activation delays and deficiencies identified in this study, doctors can optimize recovery and functional outcomes for individuals with CLAI (Watabe et al., 2021). Such strategies not only aim to restore normal muscle activation patterns but also to prevent compensatory mechanisms that could lead to secondary injuries.

This study also has several limitations. Firstly, only one patient with stabilized chronic lateral ankle instability was included in the development of the finite element model. Given inherent individual differences, the conclusions drawn from the study may vary. Additionally, while the ligaments were modeled as hyperelastic in this study, the overall rigidity of the model was only tailored to a single individual and did not fully encompass collective variations. Moreover, the boundary conditions of the models were uniform and did not simulate the actual process of landing injury. Variations in setup conditions such as material properties, mesh size, and mesh behavior could significantly influence the results. It's important to acknowledge that this scenario may not entirely reflect real-world conditions. Meanwhile, only displacement of navicular height was used in the model validation. Subsequent experiments should include validation through plantar pressure distribution, joint contact stress, and contact area.

6 Conclusion

In this study, we observed the impact of lateral ankle ligament laxity on the biomechanical characteristics and stability of the foot. The results indicate that injury or laxity of the lateral ligaments may lead to increased stress on the metatarsals, thereby compromising the stability of the ankle joint. Specifically, further laxity of the ATFL and CFL significantly altered the distribution of stress on the metatarsals, resulting in a higher risk of ankle injury. Additionally, we observed that ligament laxity may trigger muscle compensation, further affecting the stability of the foot structure. These findings are crucial for understanding the rebuilding of muscle strength during the rehabilitation process following ligament injuries and for preventing recurrent injuries. Therefore, maintaining the stability of the lateral ankle ligaments is essential for preserving the function and stability of the foot joints.

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 humans were approved by Ethical committee of Ningbo University. The studies were conducted in accordance with the local legislation and institutional requirements. The participants provided their written informed consent to participate in this study.

Author contributions

ZZ: Conceptualization, Data curation, Formal Analysis, Investigation, Methodology, Writing–original draft. HZ: Data curation, Investigation, Project administration, Writing–review and editing. TJ: Investigation, Methodology, Writing–original draft. DX: Data curation, Formal Analysis, Investigation, Writing–review and editing. E-CT: Data curation, Formal Analysis, Investigation, Writing–review and editing. MW: Data curation, Formal Analysis, Writing–review and editing. YG: Conceptualization, Formal Analysis, Funding acquisition, Project administration, Resources, Writing–review and editing.

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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 author(s) declared that they were an editorial board member of Frontiers, at the time of submission. This had no impact on the peer review process and the final decision.

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Single-belt vs. split-belt treadmill symmetry training: is there a perfect choice for gait rehabilitation post-stroke?

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Post-stroke gait asymmetry leads to inefficient gait and a higher fall risk, often causing limited home and community ambulation. Two types of treadmills are typically used for training focused on symmetry: split-belt and single belt treadmills, but there is no consensus on which treadmill is superior to improve gait symmetry in individuals with stroke. To comprehensively determine which intervention is superior, we considered multiple spatial and temporal gait parameters (step length, stride time, swing time, and stance time) and their symmetries. Ten individuals with stroke underwent a single session of split-belt treadmill training and single belt treadmill training on separate days. The changes in step length, stride time, swing time, stance time and their respective symmetries were compared to investigate which training improves both spatiotemporal gait parameters and symmetries immediately after the intervention and after 5 min of rest. Both types of treadmill training immediately increased gait velocity (0.08 m/s faster) and shorter step length (4.15 cm longer). However, split-belt treadmill training was more effective at improving step length symmetry (improved by 27.3%) without sacrificing gait velocity or step length. However, this step length symmetry effect diminished after a 5-min rest period. Split-belt treadmill training may have some advantages over single belt treadmill training, when targeting step length symmetry. Future research should focus on comparing the long-term effects of these two types of training and examining the duration of the observed effects to provide clinically applicable information.

KEYWORDS

treadmill, split-belt treadmill, gait adaptation, stroke, symmetry

1 Introduction

Stroke is a prevalent condition that affects 12.2 million people annually worldwide (Feigin et al., 2022). Over 80% of these individuals often experience deficits in walking ability, which impedes quality of life (Duncan et al., 2005; Tsao et al., 2022). Many gait rehabilitation approaches focus on improving gait velocity and spatiotemporal gait parameters, such as increasing step length or swing time, because changes in these metrics could lead to significant improvements in community walking ability and lower limb motor control (Bijleveld-Uitman et al., 2013). However, simply increasing walking speed or changing spatiotemporal parameters of both legs does not necessarily result in a more symmetric walking pattern. The person may walk faster but with an asymmetric walking pattern (Hsu et al., 2003; Wang et al., 2020). Gait asymmetry can be attributed to motor impairments such as impaired proprioception, decreased muscle strength, spasticity, and impaired balance control (Balasubramanian et al., 2007; Patterson et al., 2008; Titianova et al., 2008). The high prevalence of gait asymmetry after stroke has been linked to a higher risk of multiple negative consequences including impaired balance (Hendrickson et al., 2014), inefficient and increased metabolic costs (Ellis et al., 2013), and long-term musculoskeletal dysfunction (Jørgensen et al., 2000). Therefore, achieving spatiotemporal symmetry in gait is also crucial, as it facilitates inter-limb coordination, improves efficiency, and reduces the risk of falls (Wonsetler and Bowden, 2017).

To target gait symmetry in rehabilitation, the most common interventions are split-belt treadmill training and single-belt treadmill training, combined with visual and/or audio cueing. Split-belt treadmill training involves walking on a treadmill with belts moving at different speeds, which increases the stance time on the slow belt and swing time on the fast belt. This creates a motor error that reorganizes gait patterns, which are then applied when returning to tied-belt or overground walking (Hinton et al., 2020). Multiple studies have demonstrated that a single session of split-belt treadmill training instantly improves step length symmetry (Reisman et al., 2009; Lauzière et al., 2014; Malone and Bastian, 2014; Hirata et al., 2019). Similarly, studies have shown that combining single-belt treadmill training with visual or auditory cueing can improve gait symmetry in individuals post-stroke (Roerdink et al., 2007; Mainka et al., 2018; Shin and Chung, 2022). The visual or auditory cueing provides closed-loop sensory feedback, enabling real-time adjustments and improving the effectiveness of the training (Baram, 2013). While both interventions effectively improve gait symmetry, they also pose unique concerns. Reisman et al. (2013) observed that 12 training sessions of split-belt treadmill training improved only step length symmetry, but not temporal symmetries or gait velocity. Absence of control groups in prior split-belt studies further limits the evidence for the effectiveness of this training method. Although single-belt training is much more accessible and can successfully increase gait velocity, it does not enhance symmetry, unless combined with therapist cueing. Moreover, the evidence for using a single-belt combined with cueing to improve symmetry is far less established than split-belt treadmill training (Roerdink et al., 2007; Mainka et al., 2018; Shin and Chung, 2022). However, to date, there is no comparative analysis between split-belt and single-belt training combined with cueing interventions to establish which intervention is more effective for improving gait velocity, spatiotemporal parameters, and symmetry.

Therefore, this study aimed to 1) compare the acute effects of both interventions on spatiotemporal parameters and symmetry immediately after a single session of training and 2) determine whether the changes persisted after a 5-min period of seated rest, and 3) assess the amounts of symmetry improvement relative to the baseline symmetry for each treadmill training type. We hypothesized that both interventions would impact spatiotemporal parameters, but that split-belt training would improve step length symmetry more. Additionally, we predicted the 5-min rest period would reduce training effects since the newly-adopted movement pattern might diminish without reinforcement.

2 Materials and methods

2.1 Subjects

Ten participants with chronic stroke were recruited (Table 1). Inclusion criteria for the participants were: 1) history of a single, unilateral, supratentorial, stroke at least 1 year prior to participation 2) comfortable gait speed less than 1.0 m/s, and 3) medically stable with medical clearance to participate (absence of concurrent illness, including unhealed bone fractures or pressure sores, active injuries or infections, cardiopulmonary disease, osteoporosis, peripheral nerve damage in the lower limbs, and a history of any neurologic conditions). Exclusion criteria were as follows: 1) history of multiple strokes or bilateral strokes, 2) pregnant or nursing, 3) Modified Ashworth Score of three or greater in the lower extremity muscle groups, 4) Botox injections in the lower extremity within the last 4 months, or 5) presence of severe contractures in the lower extremities. All participants gave informed consent before participation. All participants gave informed consent before participation. All study-related procedures were approved by the Northwestern University Institutional Review Board (STU00215009), Northwestern University, Chicago, IL, United States. The study protocol was registered at clinicaltrials.gov (NCT05167786).

2.2 Protocol

Participants completed two interventions on separate days in a random order: one single-belt treadmill (Woodway® United States, Waukesha, WI) session and one split-belt treadmill (Woodway® United States, Waukesha, WI) session. Interventions started with a baseline gait assessment followed by 30-min of treadmill training. Participants were reassessed immediately after training and again after 5 min of seated recovery (Figure 1). Assessments included three 10-m walk tests at self-selected (SSV) and three 10-m walk test at fast velocity (FV) over a GAITRite® walkway (CIR Systems Inc., NJ, United States). For FV, participants were instructed to walk as fast and safely as possible. Baseline trials were processed to 1) calculate average gait velocities and 2) determine which leg had a shorter step length.

The split-belt treadmill (Woodway® United States, Waukesha, WI) training started with 2 min of tied-belt walking (both belts at ½ FV), 25 min of split-belt walking (fast belt at FV, slow belt at ½FV), and finally 5 min of tied-belt walking (goal: both belts at ½ FV). The leg with shorter step length was placed on the fast belt. Single-belt treadmill training (Woodway® United States, Waukesha, WI) consisted of 2 min

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TABLE 1 Demographic and clinical characteristics of participants.

Subject	Sex	Age (years)	Height (cm)	Weight (kg)	Time since stroke (years)	Ankle-foot orthosis	Comfortable walking speed (m/s)	Fast walking speed (m/s)
S1	F	58	162.6	85.7	4.8	L articulated	0.84	1.13
S2	M	55	175.3	117.9	8.8	L carbon fiber	0.97	1.42
S3	M	69	172.7	77.1	7.5	L articulated	0.48	0.55
S4	M	61	172.7	77.1	4.8	R articulated	0.73	0.88
S5	F	49	157.5	99.8	6.5	None	0.91	1.26
S6	M	55	165.1	72.6	7.6	None	0.88	1.14
S7	M	57	175.3	70.3	8.6	R solid	0.73	0.90
S8	M	53	175.3	79.4	2.2	None	0.90	1.19
S9	F	62	160.0	72.6	7.6	None	0.93	1.19
S10	F	73	162.6	65.8	2.7	None	0.73	0.82
Mean (SD)	4F 6M	59.2 (6.9)	167.9 (6.7)	81.8 (15.0)	6.1 (2.2)	NA	0.8 (0.1)	1.0 (0.2)

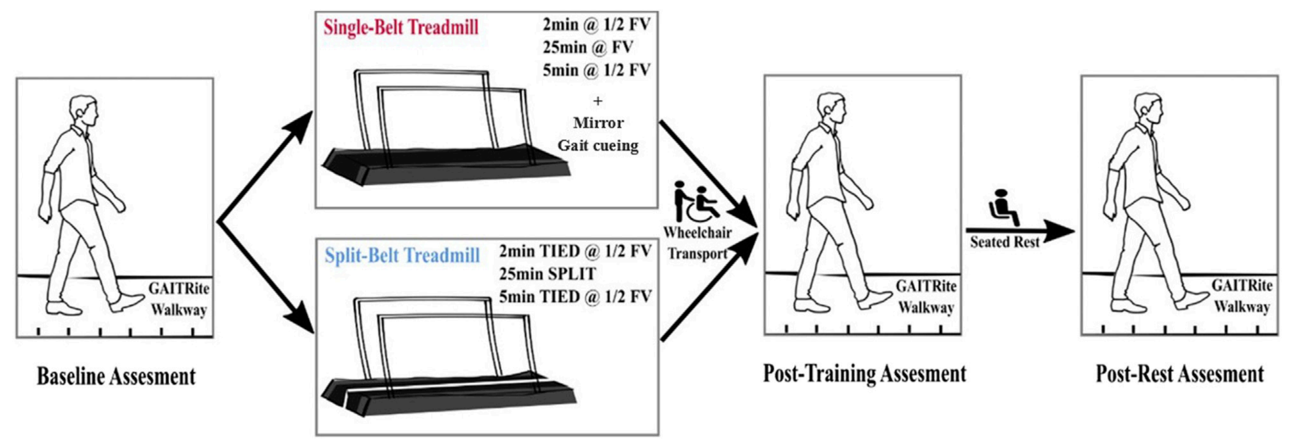

FIGURE 1
Schematic protocol of the study. During the split-belt walking, "TIED" or "SPLIT" indicate when two belts moved at the same speed or different speeds, respectively. The leg with the shorter step length was on the fast belt at FV, the leg with the longer step length was on the slow belt at ½FV.

warm up (½ fast over-ground speed), 25 min of training (fast over-ground speed), and 5 min of cool down (½ fast over-ground speed). During single-belt training, a physical therapist cued for spatiotemporal symmetry and a mirror was placed in front of the treadmill for visual feedback. Before the first split-belt training session, each participant underwent a familiarization period on the treadmill to get used to walking on the device. The participants were encouraged not to hold on to the handrails during both sessions. A ceiling-mounted safety harness was utilized without providing body weight support. Sitting breaks were provided upon participant's request. Blood pressure (BP) was assessed pre and post ambulation. Heart rate (HR) and Rated Perceived Exertion (RPE) were monitored throughout the training. Participants were transported *via* wheelchair to complete post-training assessments to avoid walking between training and assessment.

2.3 Data collection and analysis

During each assessment, step length, stride length, swing time, and stance time of over-ground gait were collected by the GAITRite walkway. Spatiotemporal symmetry indices were calculated (Eq. 1) for each participant from the obtained metrics.

$$Symmetry\ Index = \frac{|X_{shorter} - X_{longer}|}{0.5(X_{shorter} + X_{longer})} \tag{1}$$

$X_{shorter}$ and X_{longer} are the value of each spatiotemporal parameter on the shorter and longer side, respectively. The spatiotemporal parameters and symmetry values were calculated for SSV at baseline, post-training (immediate change), and post 5 min of rest (delayed change). A smaller symmetry index value indicates a more symmetric gait. Spatiotemporal symmetries were used to obtain

TABLE 2 Gait parameters in self-selected velocity. Values are shown as mean ± standard error in the table. * indicates significant difference.

	Single-belt treadmill			Split-belt treadmill			*p*-value		
	Baseline	Post-training	Post-rest	Baseline	Post-training	Post-rest	Treadmill effect	Time effect	Treadmill * time interaction
Spatiotemporal parameters									
Gait velocity (m/s)	0.78 ± 0.07	0.82 ± 0.08	0.84 ± 0.06	0.80 ± 0.06	0.82 ± 0.07	0.89 ± 0.07	0.218	*<0.001	0.271
Longer step length (cm)	55.06 ± 3.35	57.58 ± 4.22	59.03 ± 3.36	57.77 ± 3.82	57.39 ± 3.95	59.78 ± 3.69	0.300	*0.001	0.103
Shorter step length (cm)	44.39 ± 3.38	47.24 ± 3.47	49.06 ± 2.91	46.68 ± 3.45	49.66 ± 3.75	50.31 ± 3.35	*0.033	*0.002	0.502
Longer stride length (cm)	99.91 ± 6.22	105.28 ± 7.53	108.63 ± 6.07	104.73 ± 7.02	107.65 ± 7.61	110.63 ± 6.79	0.113	*0.002	0.557
Shorter stride length (cm)	99.11 ± 6.15	104.67 ± 7.50	107.67 ± 5.97	103.99 ± 6.96	106.96 ± 7.58	109.57 ± 6.69	0.103	*0.002	0.563
Longer stance time (s)	0.99 ± 0.07	0.99 ± 0.09	0.95 ± 0.07	0.98 ± 0.08	0.97 ± 0.09	0.94 ± 0.07	0.234	*0.027	0.913
Shorter stance time (s)	0.84 ± 0.06	0.85 ± 0.08	0.80 ± 0.06	0.83 ± 0.07	0.82 ± 0.08	0.79 ± 0.06	0.248	*0.011	0.797
Longer swing time (s)	0.51 ± 0.03	0.49 ± 0.03	0.50 ± 0.03	0.50 ± 0.03	0.50 ± 0.02	0.50 ± 0.02	0.765	0.193	0.371
Shorter swing time (s)	0.35 ± 0.01	0.35 ± 0.01	0.36 ± 0.01	0.35 ± 0.01	0.35 ± 0.01	0.35 ± 0.01	0.359	0.839	0.943
Symmetry parameters									
Step length symmetry	0.23 ± 0.06	0.21 ± 0.03	0.19 ± 0.03	0.22 ± 0.04	0.16 ± 0.02	0.18 ± 0.03	*<0.001	0.069	*0.005
Stride length symmetry	0.03 ± 0.01	0.03 ± 0.01	0.03 ± 0.01	0.03 ± 0.01	0.04 ± 0.01	0.03 ± 0.01	0.623	0.167	0.484
Stance time symmetry	0.17 ± 0.02	0.16 ± 0.02	0.17 ± 0.02	0.17 ± 0.02	0.17 ± 0.02	0.17 ± 0.02	0.548	0.981	0.542
Swing time symmetry	0.35 ± 0.04	0.33 ± 0.05	0.33 ± 0.03	0.35 ± 0.04	0.35 ± 0.03	0.33 ± 0.03)	0.760	0.545	0.552

the percent changes from baseline as shown in Eqs 2, 3. A positive percent change indicates a greater post value compared to baseline or increased asymmetry.

$$Immediate\ change = \frac{\left(V_{Post-training} - V_{Baseline}\right)}{V_{Baseline}} \times 100\% \tag{2}$$

$$Delayed\ change = \frac{\left(V_{Post-rest} - V_{Baseline}\right)}{V_{Baseline}} \times 100\% \tag{3}$$

2.4 Statistical analysis

The Kolmogorov–Smirnov test was used to assess the normality of the data. The data of all the outcomes was normally distributed. Generalized estimating equations (GEEs; Link Function = Identity; Structure of Covariance Matrix = Exchangeable) were conducted to test the effects of treadmill intervention (single vs. split belt) and time (baseline vs. post-training vs. post-rest) on all gait spatiotemporal and symmetry parameters. An alpha level was set at 0.05 *a priori*. Interaction effects were examined by *post hoc* pairwise comparisons with sequential Bonferroni adjustments. GEE was selected because it obtains higher power with a small sample size compared to repeated measured analysis of variance (Ma et al., 2012; Naseri et al., 2016). Spearman correlations were calculated to test the relationship between the symmetry changes (Immediate/Delayed change) and baseline gait parameters. Statistics were performed in SPSS (SPSS Statistics v27, IBM Corp., US).

3 Results

3.1 Spatiotemporal gait parameters

Statistical results are in Table 2. We found significant changes with time for gait velocity ($X^2 = 21.45$, $p < 0.001$), shorter ($X^2 = 4.55$, $p = 0.033$) and longer ($X^2 = 13.27$, $p = 0.001$) step lengths, shorter ($X^2 = 12.76$, $p = 0.002$) and longer ($X^2 = 12.87$, $p = 0.002$) stride lengths, and shorter stance time ($X^2 = 9.03$, $p = 0.011$). Pairwise comparisons showed gait velocity improved (10.13%

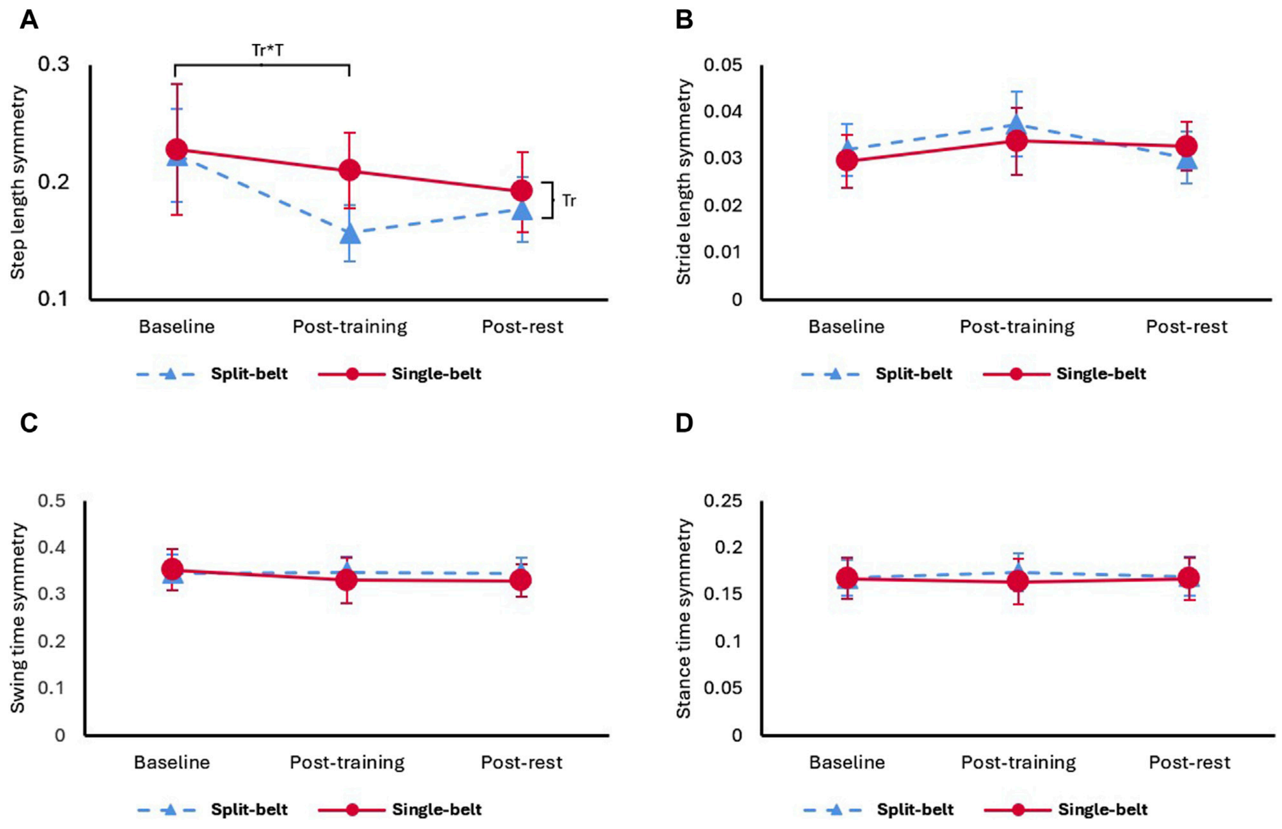

FIGURE 2
(A) Step length symmetry, **(B)** stride length symmetry, **(C)** swing time symmetry, and **(D)** stance time symmetry under self-selected velocity in two treadmill training. Tr: main effect of treadmill, Tr*T: interaction effect of treadmill and time.

increase, $p < 0.001$) after rest compared to baseline for both training types. In both training types, shorter step length increased immediately after training (6.41% increase, $p = 0.004$). Longer step length (5.32% increase, $p = 0.004$), shorter stride length (6.96% increase, $p = 0.001$) and longer stride length (7.14% increase, $p = 0.001$) all increased after rest compared to baseline. Shorter stance time (3.61% decrease, $p = 0.044$) became even shorter after rest.

3.2 Symmetry

Only step length symmetry showed a significant treadmill*time interaction ($X^2 = 10.51$, $p = 0.005$) and a significant treadmill effect ($X^2 = 11.40$, $p < 0.001$), all other symmetry parameters did not change with time or treadmill. Figure 2 shows step length symmetry immediately improved only after split-belt (27.27% decrease, $p = 0.040$), but not single-belt treadmill training ($p = 0.509$). However, after rest, there was no significant difference between the two treadmill training types.

3.3 Correlations

The immediate change of symmetry was not statistically correlated with any baseline gait parameters for either treadmill training type. However, the delayed change of SSV stride length symmetry ($R = -0.661$, $p = 0.038$; Figure 3B) and swing time symmetry ($R = -0.661$, $p = 0.038$; Figure 3C) were significantly correlated with baseline symmetry in single-belt treadmill training. Greater spatiotemporal symmetry improvements were associated with worse spatiotemporal symmetry at baseline for the single-belt treadmill only.

4 Discussion

This is the first comparison of single-session effects of single-belt and split-belt treadmill training on spatiotemporal measures in stroke survivors. Our results demonstrated that: 1) Both treadmill training types immediately increased shorter step length. Step length symmetry immediately improved significantly only after split-belt treadmill training, without compromising temporal symmetry or gait velocity. 2) Post-training rest of 5 min improved gait velocity and spatial gait performance. However, unlike the gait velocity and spatial gait performance, step length symmetry was insignificantly different from baseline after the rest. 3) Swing time and stride length symmetry improvements were associated with poor baseline levels in single-belt but not associated with baseline levels in split-belt training. These findings suggest that split-belt treadmill training might be superior to single-belt treadmill training when

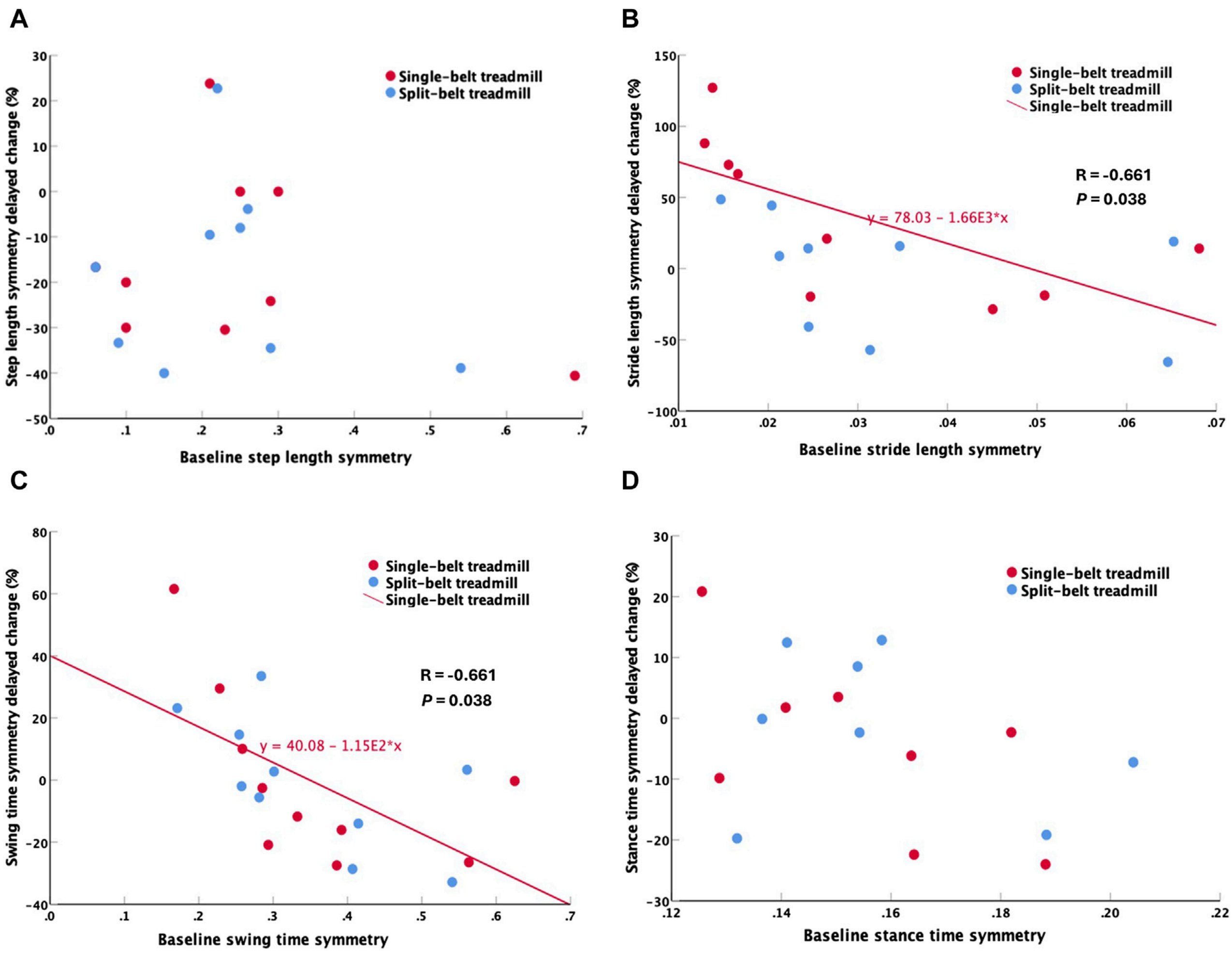

FIGURE 3
Correlations between baseline and delayed change. **(A)** Step length symmetry, **(B)** Stride length symmetry, **(C)** Swing time symmetry, **(D)** Stance time symmetry. Stride length symmetry and swing time symmetry showed significant correlations between baseline and delayed change only in single-belt treadmill training.

specifically targeting step length symmetry. However, both types of training were found to improve gait velocity and shorter step length. More research is needed to compare the long-term effects since the step length symmetry effects diminish after 5 min of rest.

4.1 Immediate training effects

Shorter step length was the only spatiotemporal parameter to show immediate training effects. The shorter step length immediately increased after both treadmill training, as observed in previous studies (Gama et al., 2017; Betschart et al., 2018). This could be attributed to the increased range of motion of the limb at faster velocities. Our split-belt training protocol placed the shorter step length side on the fast belt, which increased the step length on that side while maintaining the longer step length on the slow belt, leading to immediate improvement in symmetry. In contrast, with single-belt training, step length increased on both sides, thus not altering step length symmetry.

4.2 Rest effects

We hypothesized that the training effects on spatiotemporal gait parameters would diminish after 5 min of rest, but our results showed the opposite. The rest reduced fatigue, which possibly amplified the training effect. Reisman et al. (2013) reported no gait velocity improvement after 12 split-belt training sessions. This might also be attributed to fatigue that was induced by their longer training duration at each session. Stroke survivors adapt to split-belt training slower than neurologically intact controls do (Savin et al., 2013; Tyrell et al., 2014). To develop an effective split-belt training paradigm, future studies could test various training and rest durations.

4.3 Differences between treadmills

In this study, only split-belt treadmill training immediately improved step length symmetry without compromising gait velocity or other symmetry parameters. Additionally, single-belt treadmill baseline values were negatively correlated with the spatiotemporal symmetry improvement. Therefore, single-belt treadmill training requires worse baseline symmetry in stride length and swing time to generate spatiotemporal symmetry improvements, while this was not necessary for split-belt treadmill training. This supports choosing split-belt treadmill training to target symmetry in stroke rehabilitation. Using the error augmentation strategy (Reisman et al., 2013), we placed the leg with shorter step length on the fast belt, the fast belt further shortened the step length of the leg, exaggerating the "error" of step length asymmetry. Afterwards, when assessing the effects on gait, the aftereffects led to participants correcting "the error", resulting in the observed increased step length in the short side and maintenance of step length in longer side which in turn improved step length symmetry (Helm and Reisman, 2015). Split-belt training induces proprioceptive feedback through walking in an abnormal pattern, which informs the central pattern generators and supraspinal centers to modify the motor output to adapt and achieve a new gait pattern (Hinton et al., 2020). In contrast, single-belt treadmill walking maintains a pattern similar to over ground walking, and the participants correct the walking pattern according to the visual or auditory feedback (Pereira et al., 2016), where we observe an increase on step length symmetry by increasing both the short and long step lengths, although the improvements on symmetry are not significant.

4.4 Clinical implications and future research directions

Our study suggests that both single-belt and split-belt treadmill training effectively improve gait speed and step length on the shorter side in individuals with asymmetrical gait patterns. More interestingly, temporal symmetry remained unchanged after split-belt treadmill training. Our results indicate that split-belt treadmill training improves step length symmetry without compromising temporal symmetry, aligning with findings from a previous study by Lewek et al. (2018). Clinicians should incorporate split-belt treadmill training to target step length symmetry and consider additional strategies to maintain these improvements. This study is limited by its small sample size and its use of single training session results. To provide better suggestions for clinicians, future studies could consider increasing the sample size and conducting multiple sessions of training. Long-term effects of both training types should be investigated to understand the sustainability of improvements.

5 Conclusion

Our results demonstrate that both single-belt and split-belt treadmill training equally improve gait speed and step length on the shorter side. Split-belt training resulted in a significant improvement in step length symmetry immediately after training without impairing other temporal symmetries. However, this effect diminished after a 5-min rest. Interestingly, the short period of post-training rest reinforced spatial gait improvements from both types of treadmill training, which might be a result of reduced fatigue. However, further studies are needed to explore the long-term training effects between different types of treadmill, as the step length symmetry tends to converge between two treadmill training after a 5-min rest period. These findings highlight the potential of split-belt treadmill training to enhance gait symmetry in stroke rehabilitation. By refining and extending these training protocols, we have the opportunity to significantly improve patient outcomes, leading to more efficient and safer ambulation for individuals post-stroke.

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 humans were approved by Northwestern University Institutional Review Board (STU00215009). The studies were conducted in accordance with the local legislation and institutional requirements. The participants provided their written informed consent to participate in this study.

Author contributions

CY: Writing–review and editing, Writing–original draft, Visualization, Project administration, Methodology, Investigation, Formal Analysis, Conceptualization. NV: Writing–review and editing, Validation, Methodology, Investigation, Conceptualization. KM: Writing–review and editing, Validation, Methodology, Data curation. SA: Writing–review and editing, Validation, Methodology, Data curation. KE: Writing–review and editing, Validation, Supervision, Project administration. AK: Writing–review and editing, Validation. ER: Writing–review and editing, Supervision. AJ: Writing–review and editing, Supervision, Resources, Project administration, Funding acquisition.

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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.

Publisher's note

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Effects of transcranial direct current stimulation combined with Bosu ball training on the injury potential during drop landing in people with chronic ankle instability

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Purpose: To investigate the effects of transcranial direct current stimulation (tDCS) combined with Bosu ball training on the injury potential during drop landing in people with chronic ankle instability (CAI).

Methods: A total of 40 participants with CAI were recruited and randomly divided into the tDCS + Bosu and Bosu groups. The people in the tDCS + Bosu group received intervention of tDCS combined with Bosu ball training, and those in the Bosu group received intervention of sham tDCS and Bosu ball training, for 6 weeks with three 20-min sessions per week. Before ($week_0$) and after ($week_7$) the intervention, all participants drop-landed on a trap-door device, with their affected limbs on a moveable platform, which could be flipped 24° inward and 15° forward to mimic an ankle inversion condition. The kinematic data were captured using a twelve-camera motion capture system. Two-way ANOVA with repeated measures was used to analyze data.

Results: Significant group-by-intervention interactions were detected in the peak ankle inversion angular velocity (p = 0.047, η^2_p = 0.118), the time to peak ankle inversion (p = 0.030, η^2_p = 0.139), and the plantarflexion angle at the moment of peak ankle inversion (p = 0.014, η^2_p = 0.173). Post hoc comparisons showed that compared with $week_0$, the peak ankle inversion angular velocity and the plantarflexion angle at the moment of peak ankle inversion were reduced, the time to peak ankle inversion was advanced in both groups at $week_7$, and the changes were greater in the tDCS + Bosu group compared to the Bosu group. And, a significant intervention main effect was detected in the peak ankle inversion angle in the two groups (p < 0.001, η^2_p = 0.337).

Conclusion: Compared with the Bosu ball training, the tDCS combined with Bosu ball training was more effective in reducing the injury potential during drop landing in people with CAI.

KEYWORDS

chronic ankle instability, landing, ankle sprain, transcranial direct current stimulation, Bosu ball training

1 Introduction

Ankle sprains are one of the most common sports injuries (Fong et al., 2007), accounting for approximately 40% of all sports injuries (Colville, 1998), with recurrence rates as high as 70%–80% (McKay et al., 2001). Approximately 40% of people experienced ankle sprains developed to chronic ankle instability (CAI) (Doherty et al., 2016; Hertel and Corbett, 2019). Those with CAI are characterized by recurrent ankle sprains with persistent symptoms such as pain, recurrent episodes, swelling, limited motion, weakness, and self-reported functional impairment (Hertel and Corbett, 2019). The recurrent sprains make people with CAI prone to experience long-term degenerative sequelae such as post-traumatic ankle osteoarthritis (Hintermann et al., 2002), reduced physical activity levels (Hubbard-Turner and Turner, 2015), and decreased health-related quality of life (Arnold et al., 2011). In the United States, approximately 2 million people suffer from acute ankle sprains each year (Feger et al., 2017), with a medical cost of about $6.2 billion (Gribble et al., 2016).

Drop landing is a common maneuver in sports activities requiring strong dynamic stability and a common scenario for lateral ankle sprains (LAS) due to excessive ankle inversion (Doherty et al., 2014). Ankle sprains occur mostly on the lateral side at a rate of up to 90% (McKay et al., 2001), and approximately 75% of LAS occur during landing (Taghavi Asl et al., 2022). Ankle joints are easy to be inverted and/or plantarflexed during landing (Li et al., 2019; Kim et al., 2021; Stotz et al., 2021), and arthrogenic muscle inhibition of the peroneal muscles is observed in people with CAI, which can lead to decreased peroneal muscles strength (Dong et al., 2024), and ankle sprains are prone to occur when the peroneal muscles fail to resist ankle inversion in time (Jeon et al., 2021).

Ankle inversion angle and angular velocity, time to peak ankle inversion, and plantarflexion angle at the moment of peak ankle inversion are key indicators of the ankle sprain at landing. The peak ankle inversion angle (Simpson et al., 2022) and peak ankle inversion angular velocity (Terrier et al., 2014) were greater in people with CAI compared to those without CAI. When the ankle is accidently inverted, the distance between the talus and fibula increases (Fong et al., 2012), and the ligaments connecting the bones are stretched, which increases the potential of ligament injury (Medina McKeon and Hoch, 2019; Huang et al., 2021). Compared to those without CAI, those with CAI reached peak ankle inversion later after landing, during which time the ankle was in a state of instability and the foot was unable to adjust to the proper position to better absorb the ground reaction forces during landing (Terada and Gribble, 2015; Delahunt and Remus, 2019), resulting in the transfer of excessive ground reaction force to the joint surfaces and surrounding ligaments, increasing the potential of ankle injury and ultimately causing ankle sprain (Terada and Gribble, 2015; Delahunt and Remus, 2019). Excessive inversion and plantarflexion of the ankle during landing are the main biomechanical reasons responsible for ankle injuries (Caulfield et al., 2004). Compared to people without CAI, those with CAI landed with a greater plantarflexion angle (Delahunt et al., 2007), which increases stretching of the lateral ankle ligaments (Wright et al., 2013).

Many conventional interventions for CAI are symptom-driven, meaning that they are designed to rehabilitate deficits caused by CAI, such as strength and sensory deficits (McKeon and Donovan, 2019). However, such interventions were less effective, and people continued to experience ankle instability or experience re-injury after those interventions (McKeon and Wikstrom, 2016; Wright et al., 2017). It has been pointed out that physiologic changes in the mechanoreceptors of injured ankle ligaments or muscles are not the only factors for functional abnormalities, the maladaptive neuroplastic changes in the central nervous system (CNS), especially in the cerebral cortex, also affects the recovery of ankle function (Riemann, 2002). The adaption of CNS induces impaired sensorimotor and neurocognitive function (Needle et al., 2017; Maricot et al., 2023), and the cortical excitability decreased in the primary motor cortex (M1) in people with CAI (Needle et al., 2017; Bruce et al., 2020). In challenging condition, the inactivation of M1 leads to insufficient cognitive resources, causing abnormal movement patterns and injuries (Burcal et al., 2019; Bruce et al., 2020).

Transcranial direct current stimulation (tDCS), a noninvasive neuromodulation technique that modulates excitability and promotes neuroplasticity in the target cortex (Geiger et al., 2017), may be one of the options for CNS rehabilitation in people with CAI. tDCS has been proven to improve the excitability of the M1 and muscle activation in people with CAI (Bruce et al., 2020). Another key consideration in the use of tDCS is the selection of the motor task with which it is paired (Stagg and Nitsche, 2011), as one of its primary uses is as an adjunctive therapy to augment the acquisition of the task, i.e., motor learning. Bosu ball training can be one of the options to pair with tDCS for two reasons. Bosu balls have uneven surfaces on which instability training can be performed, and instability training has been shown to be effective for balance and postural control in people with CAI (Ha et al., 2018); And, during Bosu ball practice, participants may perform successive movements to counteract perturbations of the center of gravity, in which the CNS may learn skills continuously to counteract perturbations, and tDCS facilitates this learning process.

Therefore, this study aimed to investigate the effects of tDCS combined with Bosu ball training on the injury potential during drop landing in people with CAI, by comparing the effects of Bosu ball training only. It is hypothesized that 1. both tDCS combined with Bosu ball training and the Bosu ball training would decrease injury potential during drop landing in people with CAI, represented by the reduced peak ankle inversion angle, peak ankle inversion angular velocity, the plantarflexion angle at the moment of peak ankle inversion, and the advancement of the time to peak ankle inversion. And 2, the tDCS combined with Bosu ball training would have better effectiveness than the Bosu ball training only.

2 Materials and methods

2.1 Sample size estimate

An *a priori* power analysis (G*Power Version 3.1) indicated that a minimum of 22 participants are needed to obtain an alpha level of 0.05 and a statistical power of 0.80 based on a previous report, which compared the peak ankle inversion angular velocity in people with and without CAI when performing inversion landings under anticipated and unanticipated conditions (η^2_p = 0.064) (Han et al., 2023).

2.2 Participants

Seventy-five participants were assessed for eligibility, and 40 of them were recruited by distributing flyers at local universities. Following the guidelines of the International Ankle Consortium (Gribble et al., 2013), the inclusion and exclusion criteria were set as follows.

Inclusion criteria: Participants must have experienced at least one severe ankle sprain a year prior, causing pain, swelling, or inflammation that prevented normal participation in daily activities for more than 1 day; have had at least two episodes of ankle 'giving way' in the past 6 months; feel a persistent sense of ankle instability and functional impairment in daily activities; and score less than 24 on the Cumberland Ankle Instability Tool (Hiller et al., 2006).

Exclusion criteria: Participants with a history of lower-extremity fracture or who had undergone surgery in the past year; those who had experienced acute injuries such as lower-extremity sprains within the past 3 months; those with bilateral CAI.

The 40 participants were randomly divided into tDCS + Bosu and Bosu groups. All participants signed their approved written informed consent forms before participation. Human participation was approved by the Ethics Committee of Exercise Science of Shandong Sport University (2022043) and was in accordance with the Declaration of Helsinki.

2.3 Study design

In this single-blinded and sham-controlled study. Forty participants were randomly assigned to either the tDCS + Bosu group or the Bosu group in a 1:1 ratio. The randomization was done using a web-based randomization service (www.randomization.com). Details of the assigned group were written on cards and concealed using sequentially numbered opaque sealed envelopes. Participants in the tDCS + Bosu group underwent tDCS combined with Bosu ball training, and participants in the Bosu group underwent sham tDCS and Bosu ball training, for 6 weeks with three 20-min sessions per week. Injury potential was measured before and after the 6-week intervention.

2.4 Bosu ball training

Participants perform Bosu ball training barefoot, using a progressive training program. In weeks 1 and 2, they practiced single-leg stance, single-leg stance with forward-backward leg swing (30°–45°), single-leg stance with medial-lateral leg swing (20°–30°), and single-leg squat. In weeks 3 and 4, they practiced swallow balanced stance, single-leg stance with forward-backward leg swing (45°–60°), single leg stance with medial-lateral leg swing (30°–45°), and single-legged squat take-ups. In weeks 5 and 6, they practiced catching a ball while single-leg stance, single-leg stance with forward-backward leg swing (45°–60°), single-leg stance with medial-lateral leg swing (30°–45°), and bending over to touch the edge while single-leg stance. See illustrations of the movements in Figure 1. Each movement was performed for 30 s and repeated 5 times, with a 30-s rest between movements. The total time of each session is approximately 20 min.

2.5 tDCS intervention

tDCS was delivered by a tDCS device (StarStim8, NE transcranial direct current stimulator, Spain). Five 5 mm radius rubberized circular electrodes were used in the tDCS montage, with a central anodal electrode and four surrounding cathodal (return) electrodes. According to the 10/20 EEG template, the anode was placed at Cz, and the other four electrodes were located at Fz, Pz, C3 and C4, respectively (Villamar et al., 2013) (Figure 2A). The current intensity at the anode was set at 2 mA, and the return current was evenly distributed among the four cathodes. During tDCS, the current increased from 0 mA to 2 mA in the first 30 s, was maintained at this level for 19 min, and then gradually decreased to 0 mA in 30 s. For the sham stimulation, the same electrode location was used, but the current remained to be 0 mA during the stimulation.

2.6 Drop landing test

Before formal tests, participants wore uniform tight shorts and T-shirts. Then 36 markers were adhered to their lower limbs according to the protocol of the Oxford Foot Model (McCahill et al., 2008). Participants then had 5 min to warm up and familiarize themselves with the tests by at least 5 drop-landing trials. Then, they conducted formal drop landing test.

Participants drop-landed from a height of 30 cm (Mokhtarzadeh et al., 2017) onto a custom trap-door device consisting of three platforms, namely, take-off (Figure 2B), movable (Figure 2B), and support (Figure 2B) platforms. The device is commonly used to trigger ankle inversion during landing (Gehring et al., 2014). The surface of the movable platform would be flipped 24° inward and 15° forward when it suffered a force greater than 10 N. Each participant's affected foot landed on the movable platform and the unaffected foot landed on the support platform. A marker was placed on the lateral edge of the movable platform to identify the time point when the platform surface moved. During the drop-landing test, participants' kinematic data were recorded using a 12-camera, 3D infrared motion capture system (Vicon Vantage V5, Oxford Metrics Limited, Oxford, United Kingdom) at a frequency of 100 Hz. Participants wore a jacket with a rope that passed through a pulley on the ceiling, with the other end of the rope controlled by the tester for safety. A total of five trials were performed, and means of the five trials were calculated for further analysis.

2.7 Data processing

The data were collected from the time point at landing and 200 ms after landing (Simpson et al., 2022), which was defined by the movement of the marker placed on the lateral edge of the movable platform (Bhaskaran et al., 2015) (Figure 2B). The time stage was selected because real ankle sprain occurs within this stage (Fong et al., 2009). The raw data from the motion capture system were processed using Vicon Nexus (version: 2.10.2, Oxford Metrics, Ltd.), imported into Visual 3D software (V6 Professional, C-Motion, United States) and low-pass filtered at 10 Hz (Jackson, 1979).

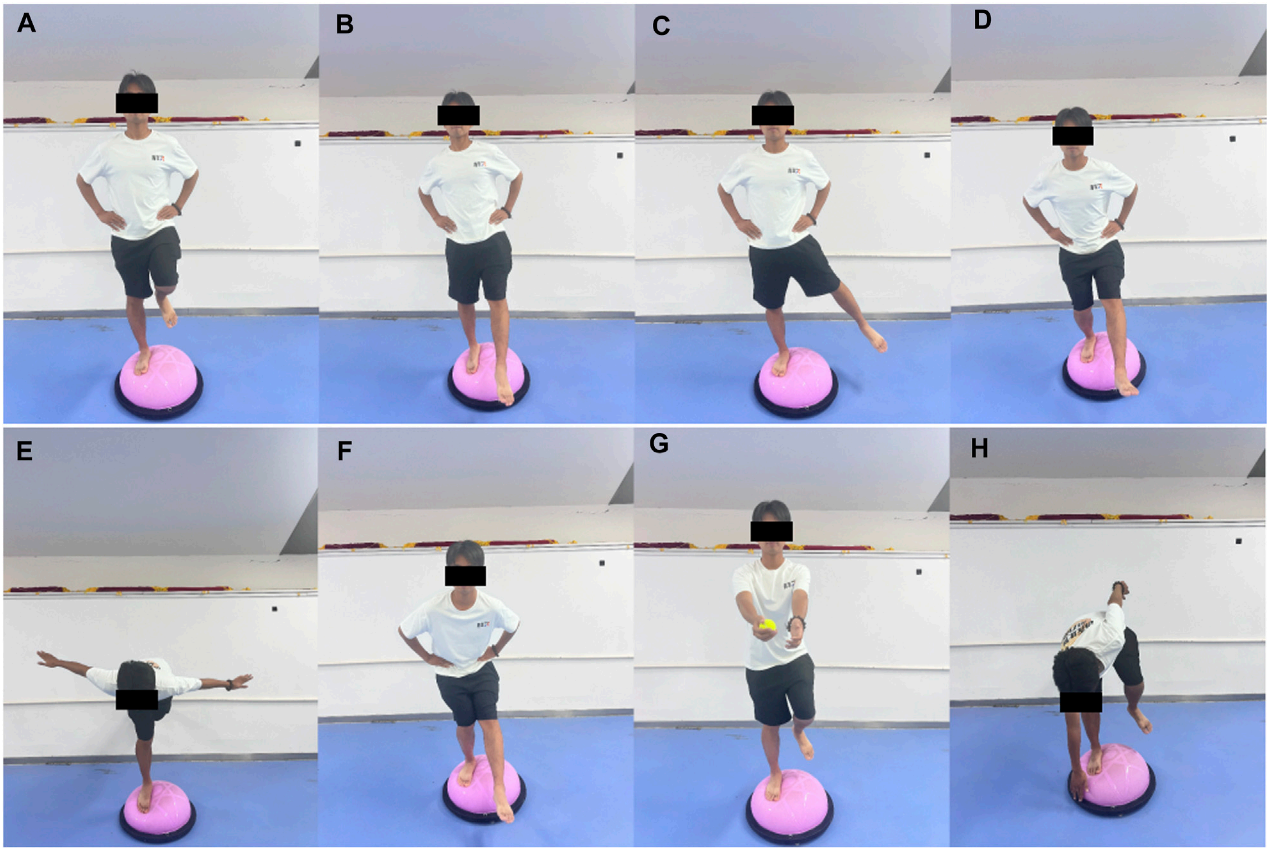

FIGURE 1
Illustrations of the Bosu ball training movements **(A)** single-leg stance, **(B)** single-leg stance with forward-backward leg swing, **(C)** single-leg stance with medial-lateral leg swing, **(D)** single-leg squat, **(E)** swallow balanced stance, **(F)** single-legged squat take-ups, **(G)** catching a ball while single-leg stance, and **(H)** bending over to touch the edge while single-leg stance.

High-frequency data, usually caused by alternating current or ground vibration, have been filtered.

2.8 Variables

The peak ankle inversion angle was defined as the maximum Euler angle of the foot relative to the tibia in the coronal plane. The peak ankle inversion angular velocity was defined as the maximum rate of changes of the ankle inversion angle, i.e., the maximum value of angular increment per unit time. The time to peak ankle inversion was defined as the time from the foot contact with the moveable platform to peak ankle inversion. And, the plantarflexion angle at the moment of peak ankle inversion was defined as the Euler angle between the line from the heel to the third metatarsal head and the tibia in the sagittal plane, at the moment of peak ankle inversion.

2.9 Statistical analysis

The normality of data was verified using Shapiro-Wilk tests. A two-way ANOVA with repeated measures was used to verify the main effects of group (tDCS + Bosu vs. Bosu) and intervention ($week_0$ vs. $week_7$), and their interactions. If a significant interaction was detected, Bonferroni adjusted post hocs would be conducted. Partial eta square (η^2_p) was used to represent the effect size of main effects and interactions. The thresholds for η^2_p were as follows: 0.01–0.06, small; 0.06–0.14, moderate; >0.14, large (Pierce CA and Aguinis, 2004). Cohen's *d* was used to represent the effect size of the *post hoc* comparison. The thresholds for Cohen's *d* were as follows: <0.20, trivial; 0.21–0.50, small; 0.51–0.80, medium; >0.81, large (Cohen, 2013). All data are expressed as mean ± standard deviation. The significance level is set to 0.05 or 5%, and *p*-value less than the level indicates a statistically significant result, meaning the observed data provide strong evidence against the null hypothesis.

3 Results

The Shapiro-Wilk test showed that all dependent variables were normally distributed. Figure 3 shows that a total of 40 participants were recruited for this experiment and were divided into two groups. During the intervention, six participants were excluded due to time commitment, and after the 6-week intervention, 18 participants remained in the tDCS + Bosu group (20.1 ± 1.3 years; 175.5 ± 8.0 cm; 72.4 ± 9.6 kg) and 16 in the Bosu group (21.0 ± 1.8 years; 173.3 ± 12.0 cm; 68.9 ± 11.5 kg). There were no significant differences in age, height, and body mass between the two groups.

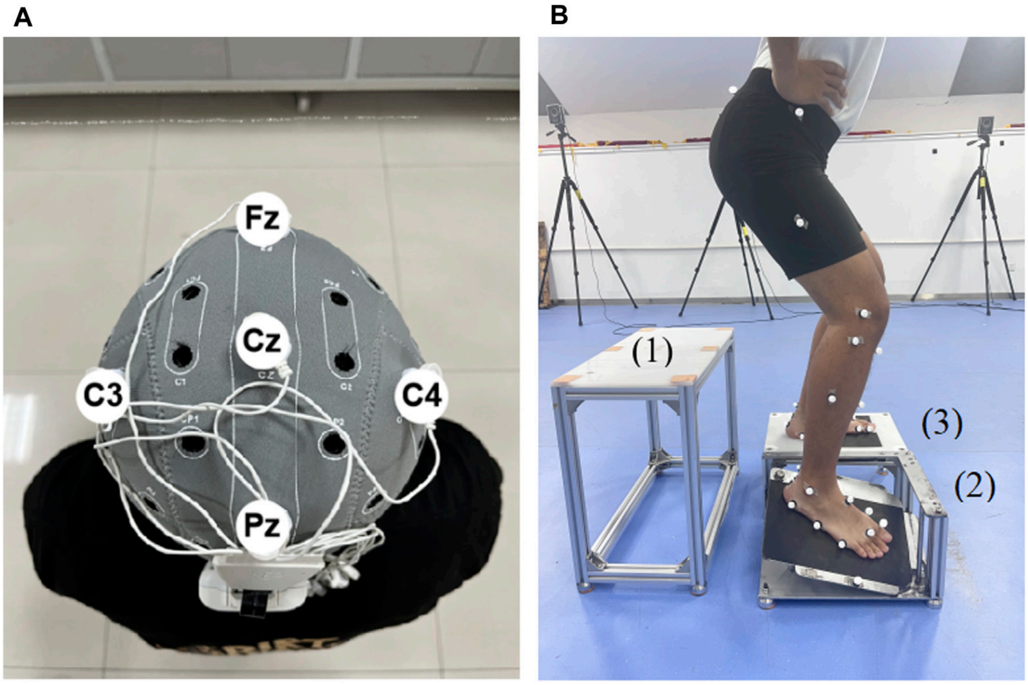

FIGURE 2
Illustration of tDCS electrode placement and drop landing test **(A)** The illustration of the tDCS electrode placement. The anode was placed over Cz of the 10/20 EEG template; the four cathodes were placed over Fz, C3, Pz, and C4. **(B)** The illustration of the drop landing test. 1, take-off platform. 2, moveable platform. 3, support platform.

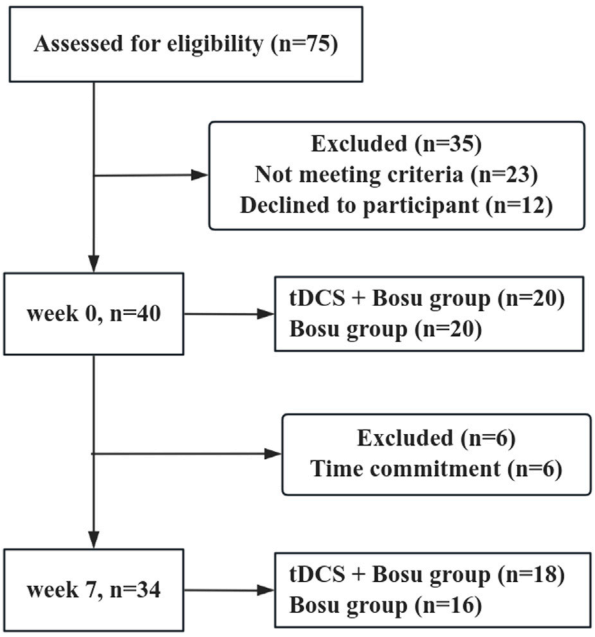

FIGURE 3
Participant flow chart. Participation flow chart from week 0 to week 7. The final analysis included data from 34 participants. Forty-one participants were excluded from the original 75 assessed due to various reasons.

Figure 4 shows significant group by intervention interactions in peak ankle inversion angular velocity ($p = 0.047$, $\eta^2_p = 0.118$), time to peak ankle inversion ($p = 0.030$, $\eta^2_p = 0.139$), and plantarflexion angle at the moment of peak ankle inversion ($p = 0.014$, $\eta^2_p = 0.173$). After 6 weeks of intervention, the ankle peak inversion angular velocity and the plantarflexion angle at the moment of peak ankle inversion were reduced in both groups, and the time to peak ankle inversion was advanced in both groups, and the changes were greater in the tDCS + Bosu group compared to the Bosu group. Moreover, a significant intervention main effect in both groups was detected in the peak ankle inversion angle ($p < 0.001$, $\eta^2_p = 0.337$).

4 Discussion

The purpose of this study was to verify the effect of tDCS combined with Bosu ball training on the injury potential during drop landing in people with CAI. These results partially support our hypotheses by pointing out that tDCS + Bosu training has better effects in reducing peak ankle inversion angular velocity and the plantarflexion angle at the moment of peak ankle inversion and advancing time to peak ankle inversion than Bosu training only, and both two interventions reduced the peak ankle inversion angle.

Our results showed that Bosu ball training as well as tDCS + Bosu ball training reduced the peak ankle inversion angle, and tDCS

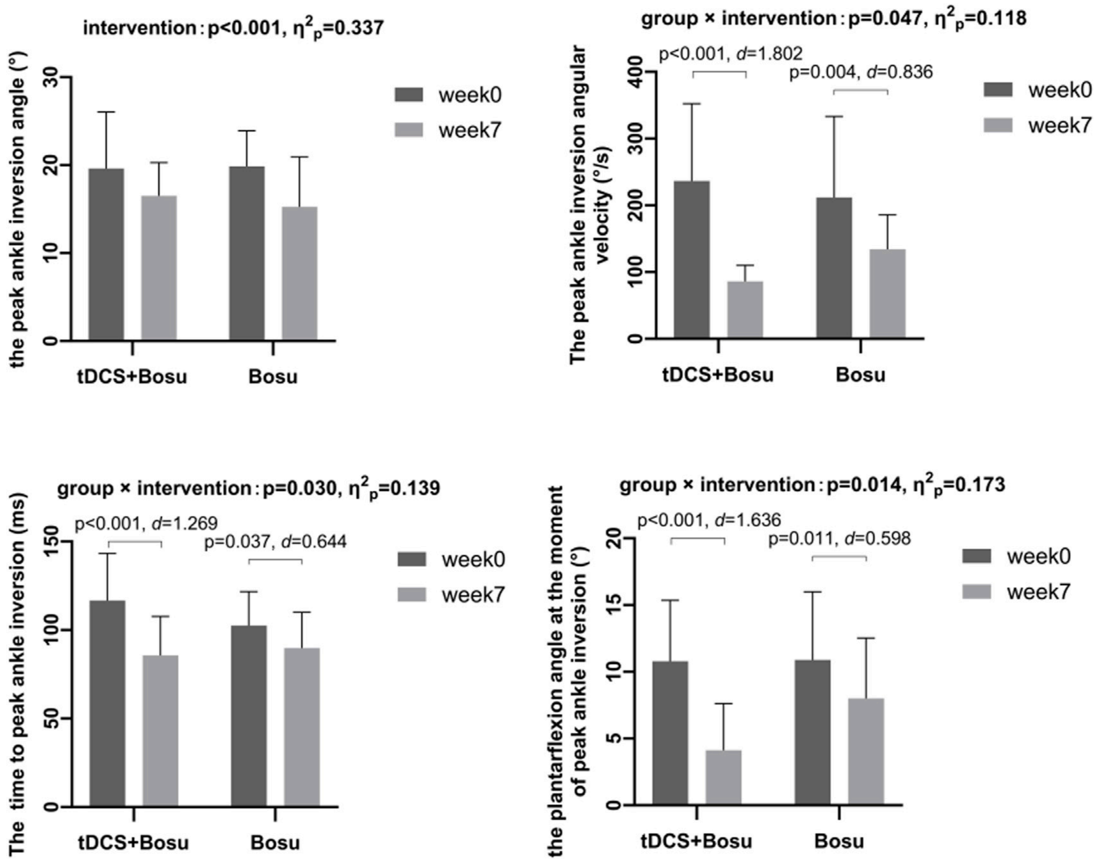

FIGURE 4
Comparison of kinematic data during drop-landing before and after interventions in people with chronic ankle instability **p* < 0.05 indicates a statistical difference.

+ Bosu ball training did not show an extra effect. It could be inferred that most of the effects were due to Bosu ball training. We believe that the effectiveness of Bosu ball training lines in the role of the unstable surface and its potential to improve balance and postural control. Previous studies support us by pointing out that compared to a stable surface, training on an unstable surface has a better effect on balance and postural control in people with CAI (Ha et al., 2018). The use of unstable surface disrupts balance, increases the sensory stimulation needed between the joints and the skin, and enhances the body's balance response (Alizamani et al., 2023). Furthermore, training on an unstable surface is believed to improve proprioception (Cug et al., 2016; Alizamani et al., 2023), which has been a deficit in people with CAI and can lead to a decrease in their postural control (Freeman et al., 1965; Liu et al., 2024). We believe that Bosu ball training improves balance and proprioception, so the participants could feel the changes in ankle movements earlier, and be able to limit the ankle inversion angle during the drop landing to decrease injury potential. The inability of tDCS to further improve proprioception may be related to the underlying pathomechanism of CAI (Ma et al., 2020). The diminished proprioception in people with CAI due in part to diminished spinal cord-mediated sensory afferent input from mechanoreceptors and/or diminished postural reflex responses (Lephart et al., 1998). tDCS selectively modulates cortical excitability and thus may have limited effects on the spinal control component of the proprioceptive system (Ma et al., 2020). In addition, it has been shown that while the application of tDCS over the affected motor cortex contributes to the recovery of lower limb motor weakness, this improvement is not sufficient to restore balance function (Chang et al., 2015).

Our study showed that tDCS + Bosu ball training had better effects than Bosu ball training in reducing injury potential during drop landing in people with CAI, which may be attributed to three reasons. Primary, tDCS may increase the activation of the peroneus longus and tibialis anterior, improve motor function, and reduce the injury potential in people with CAI. Studies have shown that increasing central nervous excitability improves muscle activation (Needle et al., 2017). It has been further pointed out that tDCS on M1 increased activation of peroneus longus (Bruce et al., 2020) and tibialis anterior (Zhan et al., 2023) in people with CAI. The peroneus longus and tibialis anterior are the main ankle eversion and dorsiflexion muscles, respectively, and increased activation of them reduces the ankle inversion and plantarflexion (Kim et al., 2003) as well as the inversion and plantarflexion moments during landing in patients with CAI (Li et al., 2018), which in turn reduces the ankle inversion angular velocity and plantarflexion angle, advances the time to peak ankle inversion, and thus reduces the potential for ankle injury.

Secondary, tDCS may improve the excitability of the M1 projected to lower limb muscles. It is believed that the M1, projected to peroneus longus and tibialis anterior, is smaller in size and less excitable (Pietrosimone and Gribble, 2012; Nanbancha et al., 2019) in people with CAI compared to those without CAI. A previous study pointed out that after tDCS intervention, stroke patients showed shorter latencies and higher motor evoked potential amplitudes in the tibialis anterior, compared to those receiving sham tDCS, suggesting that tDCS increases the excitability of corticospinal tract projected to the tibialis anterior (Chang et al., 2015). Another study further indicated that tDCS promoted M1 excitability and further enhanced the function of the muscles to which it projects (Sanes and Donoghue, 2000). In our study, tDCS may have facilitated the excitability of the M1 and further enhanced the function of peroneus longus and tibialis anterior, resisted the inversion and plantarflexion during trap-door drop landing, and then reduced the ankle sprain injury potential.

Tertiary, tDCS may enhance the effects of Bosu ball training by facilitating motor learning. Recent paradigm shifts in the etiology of ankle instability have revealed changes within the CNS that alter motor planning and produce movement patterns that re-injure individuals (Needle et al., 2017). A crucial consideration when using tDCS is to select a motor task to pair it with, as its primary purpose is to enhance the acquisition of a task as adjuvant therapy (Stagg and Nitsche, 2011). Previous studies supported our viewpoint by indicating that tDCS on M1 combined with the sequential sensorimotor task of sport stacking could better improve the flexibility of hands and motor performance compared to sham tDCS in healthy people (Pixa et al., 2017). It has also been shown that tDCS improved the accuracy of tracking during movements and led to greater improvements in ankle voluntary control of their paretic ankle in stroke patients than sham tDCS when practicing a tracking task for dorsiflexion-plantarflexion movements of the paralyzed ankle (Madhavan et al., 2011). We suggest that during Bosu ball training, people with CAI continually make movements to counteract perturbations of the center of gravity. During this process, the CNS continually learns the skills of counteracting perturbations, and tDCS facilitates this learning process.

To our knowledge, this study is the first to investigate the effect of tDCS combined with Bosu Ball training on the injury potential in people with CAI. It confirms that CNS-directed rehabilitation (e.g., tDCS) is an approach that can provide additional effects to conventional functional training for the prevention of ankle re-sprains in people with CAI, and provides new ideas for the clinical development of rehabilitation programs for people with CAI.

There are several limitations to this study. First, there was no follow-up after the 6-week intervention, so we were unable to determine how long the effect of the intervention on reducing the injury potential in people with CAI lasted. Second, the Bosu ball training was combined with either tDCS or sham tDCS, the individual effects of each intervention remain unknown. Third, we used four biomechanical variables to represent injury potential, however, no epidemiological studies explored the sensitivity and specificity of biomechanical predictors during drop landing to the ankle sprain occurrence. The use of statistical predictors would improve the precision of this study.

5 Conclusion

Compared with the Bosu ball intervention, the tDCS combined with Bosu ball intervention was more effective in reducing the injury potential during drop landing in people with CAI, suggesting that tDCS can be used as an effective rehabilitation approach to reduce the injury potential of ankle sprains in people with CAI.

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 humans were approved by Ethics Committee of Exercise Science of Shandong Sport University. The studies were conducted in accordance with the local legislation and institutional requirements. The 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

XH: Data curation, Writing–original draft, Writing–review and editing. HG: Data curation, Writing–review and editing. HF: Writing–review and editing, Data curation.

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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 attentional focus strategies in drop landing biomechanics of individuals with unilateral functional ankle instability

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Background: Functional Ankle Instability (FAI) is a pervasive condition that can emerge following inadequate management of lateral ankle sprains. It is hallmarked by chronic joint instability and a subsequent deterioration in physical performance. The modulation of motor patterns through attentional focus is a well-established concept in the realm of motor learning and performance optimization. However, the precise manner in which attentional focus can rehabilitate or refine movement patterns in individuals with FAI remains to be fully elucidated.

Objective: The primary aim of this study was to evaluate the impact of attentional focus strategies on the biomechanics of single-leg drop landing movements among individuals with FAI.

Methods: Eighteen males with unilateral FAI were recruited. Kinematic and kinetic data were collected using an infrared three-dimensional motion capture system and force plates. Participants performed single-leg drop landing tasks under no focus (baseline), internal focus (IF), and external focus (EF) conditions. Biomechanical characteristics, including joint angles, ground reaction forces, and leg stiffness, were assessed. A 2 × 3 [side (unstable and stable) × focus (baseline, IF, and EF)] Repeated Measures Analysis of Variance (RM-ANOVA) analyzed the effects of attentional focus on biomechanical variables in individuals with FAI.

Results: No significant interaction effects were observed in this study. At peak vertical ground reaction force (vGRF), the knee flexion angle was significantly influenced by attentional focus, with a markedly greater angle under EF compared to IF ($p < 0.001$). Additionally, at peak vGRF, the ankle joint plantarflexion angle was significantly smaller with EF than with IF ($p < 0.001$). Significant main effects of focus were found for peak vGRF and the time to reach peak vGRF, with higher peak vGRF values observed under baseline and IF conditions compared to EF ($p < 0.001$). Participants reached peak vGRF more quickly under IF ($p < 0.001$). Leg Stiffness (k_{leg}) was significantly higher under IF compared to EF ($p = 0.001$).

Conclusion: IF enhances joint stability in FAI, whereas EF promotes a conservative landing strategy with increased knee flexion, dispersing impact and minimizing joint stress. Integrating these strategies into FAI rehabilitation programs can optimize lower limb biomechanics and reduce the risk of reinjury.

KEYWORDS

functional ankle instability, attentional focus strategies, biomechanical characteristics, landing movements, motor control

1 Introduction

Ankle sprains are among the injuries most commonly sustained during sports activities (Medina McKeon and Hoch, 2019). They are especially prevalent during high-intensity actions such as leaping and drop landing, where the risk of injury is significantly increased to between 25 and 50 percent (Wang et al., 2025). If a sprained ankle is not properly managed and restored, it can lead to chronic discomfort and recurrent swelling, which are consequences that persist (Kong et al., 2023). Diminished movement control and an increased risk of recurrent injuries can result from this, potentially evolving into Functional Ankle Instability (FAI) (Cao et al., 2019). Joint stability and proprioception can be adversely affected by FAI, which in turn impairs motor performance and overall quality of life (Cain et al., 2020; Nunes et al., 2020).

In sports activities, drop landing tasks are particularly challenging and require substantial stability and coordination of the lower limbs (Meng et al., 2022). For individuals with FAI, performing drop landing tasks presents considerable challenges. This is due to impaired ankle joint stability and proprioception, which cause difficulties in controlling joint motion and force distribution, thereby consequently increasing the risk of injury (Xue et al., 2021). Previous research indicates that individuals with FAI demonstrate substantial differences between the stable and unstable limbs regarding joint angles and ground reaction forces during drop landing tasks, that are associated with motor control impairments and heightened injury risk (Zhang et al., 2014). Therefore, examining the biomechanical characteristics of individuals with FAI during drop landings, as well as the potential for enhancing lower limb stability and inherent coordination, is a primary focus in rehabilitation research.

Attention focus strategies constitute a significant concept in sports training and rehabilitation. Adjusting an individual's attention distribution can markedly influence their motor execution efficiency and stability (Dalvandpour et al., 2021). Internal focus strategies (IF) typically direct attention towards an individual's own body movements and sensations, whereas external focus strategies (EF) emphasize attention on the external environment and targets (Chua et al., 2021). Previous studies suggest that the EF strategy may diminish dependence on impaired proprioception and improve motor patterns and efficiency (An and Wulf, 2024). For individuals with FAI, employing an EF strategy significantly enhances motor control during drop landing. Moreover, the IF strategy effectively improves the internal perception of movement execution (Aiken and Becker, 2023), may support motor control and proprioception restoration in individuals with FAI, thereby improving motor performance and self-assurance. Although current research has demonstrated that attention focus strategies positively affect sports performance (Slovák et al., 2024), there remains a paucity of research regarding how modifying attention focus can optimize the biomechanics during drop landing movement of individuals with FAI.

Thus, this study aimed to examine the kinematic and kinetic attributes of the lower limbs during single-leg drop landings in individuals with unilateral FAI, under conditions of IF and EF. We advance the following hypotheses: 1) the IF and EF strategies will exert distinct influences on both the unstable and stable limbs of individuals with FAI; 2) the judicious use of focus strategies could potentially optimize the biomechanical characteristics during single-leg drop landings for individuals with FAI.

2 Methods

2.1 Participants

A total of 18 males with FAI participated in the study, and all provided informed consent. The Soochow University Ethics Committee Board approved this study (Table 1). The inclusion criteria for patients with FAI were as follows. 1) Participants must have a history of at least one unilateral ankle sprain within the past year and a sense of instability; 2) Participants should have no history of severe lower limb injury (Wu et al., 2021), except for ankle sprains; 3) CAIT score ≤24 (Hiller et al., 2007); 4) FAI symptoms are limited to the unilateral ankle joint. The exclusion criteria were as follows: 1) Participants with a history of sprains in both ankles (Wu et al., 2021); 2) Participants with acute pathological symptoms in the lower limbs; 3) Participants with a history of lower limb surgery (Wang et al., 2022); 4) Participants with congenital joint deformities; 5) Participants with a positive tilt of the talus or anterior drawer test results.

2.2 Experimental procedure

All participants wore standardized laboratory testing attire prior to the experiment to minimize the influence of external variables on the results. Participants initially engaged in adequate warm-up activities to prepare for subsequent testing. After the warm-up,

TABLE 1 Basic information of subjects ($\bar{x} \pm S$).

Item	Experimental group (n = 18)
Age (years)	23.5 ± 1.7
Height (cm)	177.9 ± 6.3
Body mass index (kg/m^2)	23.0 ± 2.0
Cumberland ankle instability tool score	18.8 ± 1.9
Unstable side (Left\Right)	(6\12)

FIGURE 1
Marker points pasting diagram.

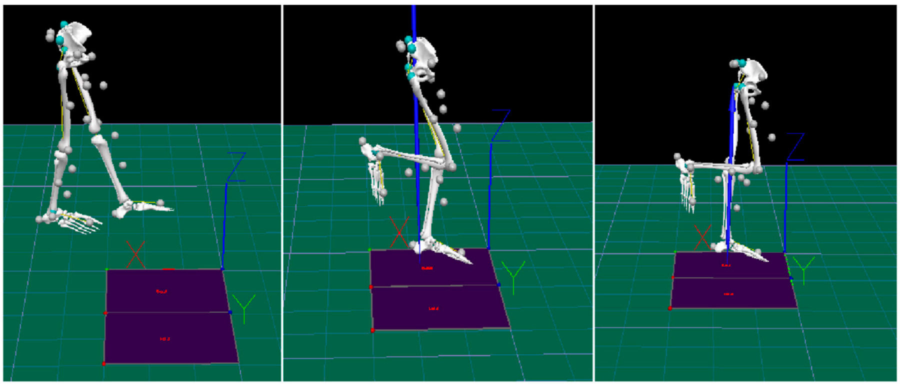

FIGURE 2
Action diagram. Participants performed a single leg drop landing from a 30-cm high platform onto a force plate.

researchers affixed 28 marker points with a diameter of 14 mm onto the participants to capture precise kinematic data (Figure 1).

Participants stood naturally on a 30 cm high jump box with feet shoulder-width apart, placing their hands on their waists during the test to prevent arm swing inertia. Following the baseline test, participants received verbal instructions for two different focus conditions.

In the IF condition, participants were instructed to focus on the action of "flexing the lower limb joints upon drop landing." In the EF condition, they were instructed to "focus on achieving a reduced impact sound upon drop landing" (Almonroeder et al., 2020).

All drop landing trials were conducted between conditions, with instructions given only once before the start of each condition to ensure consistency. Participants completed tests of single-leg drop landing tasks on both the stable and unstable sides under baseline, IF, and EF conditions. During the landing, participants were instructed to keep their non-supporting leg naturally bent to the side of their body. This posture was maintained to ensure a controlled and consistent landing motion (Figure 2). Under each test condition, participants were required to complete three successful single-leg drop landing movements, adhering to the specified landing posture. The results of the three tests for each participant were averaged to reduce random error and enhance the reliability of the data.

2.3 Data processing

Kinematic data were captured using a motion analysis system that comprised eight infrared cameras (Vicon Motion Analysis, United Kingdom) by tracking 28 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 CGM 23 lower limb 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. To enhance data quality, a fourth-order low-pass Butterworth filter was applied to smooth the three-dimensional coordinate data with a cutoff frequency set at 10 Hz; concurrently, the force platform data were filtered accordingly with a cutoff frequency of 50 Hz, and the calculation of lower limb joint angles was conducted using the Euler angle method (Yao et al., 2024). The IC was defined as the initial instance when vGRF exceeded 10 N (Christoforidou et al., 2017). The following data were analyzed (Meng et al., 2022): joint angles (°): joint angles of the hip, knee, and ankle in the sagittal and frontal planes at the initial contact (IC) and peak vertical ground reaction force (vGRF) moments; weight-normalized peak GRF [including vGRF, medial GRF (mGRF), and lateral GRF (lGRF)] (BW); Time to peak vGRF (s): This study recorded the time required for the vGRF to reach its peak from the IC; Leg Stiffness (kleg, BW/m), k_{leg} was calculated using Equation 1 (Zhang et al., 2017).

$$k_{leg} = Fz_{max} \Big/ \Delta L \tag{1}$$

Fz_{max} represents the peak vGRF, and ΔL represents the maximum length change of the lower limb during the drop landing process.

2.4 Statistical analysis

Data were analyzed using SPSS 26, with descriptive statistics presented as mean ± standard deviation ($\bar{x} \pm S$). The normality of the data was tested using the Shapiro-Wilk test to ensure the applicability of subsequent analyses. A 2 × 3 [side (stable and unstable) × focus (baseline, IF, and EF)] repeated measures analysis of variance (RM-ANOVA) with Greenhouse-Geisser correction was conducted to assess the influence of different conditions on the outcomes. The criterion for statistical significance was set at $\alpha = 0.05$. Post hoc analyses were conducted using the Bonferroni correction, with an adjusted adjusted *p*-value threshold of 0.017. Once an interaction effect be detected in the analysis, further simple effect analysis would be conducted to specify the source of the interaction effect. If no interaction effect is found, the main effects would be analyzed directly.

3 Results

At the IC, the main effect of attentional focus on the hip joint flexion angle was statistically significant ($F = 8.777$, $p = 0.001$, eta^2 = 0.403). However, *post hoc* analyses with correction for multiple comparisons did not reveal significant differences between the baseline and either the IF or EF conditions ($p > 0.017$). At the peak vGRF, a similar pattern emerged, with a significant main effect of attentional focus observed ($F = 7.401$, $p = 0.003$, eta^2 = 0.363), yet no statistically significant differences were detected after correction for multiple comparisons ($p > 0.017$) (Table 2).

At the peak vGRF, a significant main effect of attentional focus on knee flexion angle was observed ($F = 7.127$, $p = 0.012$, eta^2 = 0.354). Specifically, *post hoc* analysis with Bonferroni correction indicated that compared to the IF condition, the EF condition exhibited a significantly greater knee flexion angle ($p = 0.003$). Concurrently, a significant main effect of side was observed in knee joint varus angle, with the unstable side showing a higher varus angle compared to the stable side ($F = 6.119$, $p = 0.028$, eta^2 = 0.320) (Table 3).

At the IC, a significant main effect of side was observed on the ankle joint inversion angle, with the unstable side demonstrating a higher inversion angle compared to the stable side ($F = 15.337$, $p = 0.002$, eta^2 = 0.542); at the peak vGRF, a significant main effect of focus was noted on the ankle joint plantarflexion angle ($F = 6.369$, $p = 0.017$, eta^2 = 0.329). Post hoc analysis with Bonferroni correction indicated that the EF condition showed a significantly smaller ankle joint plantarflexion angle compared to the IF condition ($p < 0.001$) (Table 4).

Significant main effects of focus were observed on the peak vGRF and the time to peak vGRF variables ($F = 13.946$, $p < 0.001$, eta^2 = 0.518). Post hoc analysis with Bonferroni correction indicated that compared to the EF condition, participants under baseline and IF conditions exhibited a higher peak vGRF ($p = 0.002$, $p < 0.001$). Furthermore, under the IF condition, participants reached the peak vGRF in a significantly shorter time ($F = 14.936$, $p < 0.001$, eta^2 = 0.535). For the index of k_{leg}, the main effect of focus was also significant ($F = 4.859$, $p = 0.016$, eta^2 = 0.272). Post hoc analysis with Bonferroni correction found that participants under the IF condition showed greater k_{leg} compared to the EF condition ($p = 0.001$). Additionally, the main effect analysis of side indicated a significant increase in k_{leg} among participants with the unstable side ($F = 5.121$, $p = 0.041$, eta^2 = 0.283) (Table 5).

4 Discussion

4.1 The influence of attentional focus strategies on lower limb biomechanics in individuals with FAI

At the peak vGRF, individuals with FAI exhibited a significantly smaller knee flexion angle under IF conditions compared to EF conditions. The findings indicate that EF promotes a landing strategy that enhances shock absorption, beneficial for reducing impact on lower limbs. Higher knee flexion angles, associated with a "soft landing," help disperse impact forces and protect joints (Laughlin et al., 2011; Wu et al., 2013). Given that individuals with FAI may have compromised stability in their lower limb joints (Liu et al., 2024), the increased knee flexion under EF conditions acts as an effective shock absorber, contributing to the reduction of peak impact forces through extended contact time and thus maintaining movement stability—a factor critical for joint protection and injury risk reduction. (Doherty et al., 2016). Furthermore, further analysis revealed that during the drop landing process, individuals with FAI exhibited a higher varus angle at the knee joint on the unstable side, which may be a compensatory biomechanical adjustment made to adapt to the compromised stability of the ankle joint (Jamaludin et al., 2020;

TABLE 2 Hip joint angles at IC and peak vGRF moments during single drop landing.

Variables	Unstable side	Stable side	Main effect		Interaction effect
			Focus	Side	Focus × side
IC					
Hip joint flexion (+)/extension (–) angle (°)					
Baseline	16.83 ± 6.29	16.95 ± 3.95	**p = 0.001***	p = 0.555	p = 0.804
IF	23.41 ± 9.47	24.54 ± 8.23			
EF	24.04 ± 9.83	24.44 ± 7.57			
Hip joint adduction (+)/abduction (–) angle (°)					
Baseline	−4.67 ± 4.60	−6.64 ± 2.65	p = 0.170	p = 0.206	p = 0.950
IF	−6.15 ± 4.04	−7.55 ± 3.77			
EF	−6.66 ± 4.80	−8.16 ± 3.99			
Peak vGRF					
Hip joint flexion (+)/extension (–) angle (°)					
Baseline	24.04 ± 8.60	23.14 ± 6.67	**p = 0.003***	p = 0.787	p = 0.607
IF	29.49 ± 10.01	30.80 ± 9.01			
EF	33.10 ± 11.23	33.65 ± 8.65			
Hip joint adduction (+)/abduction (–) angle (°)					
Baseline	−3.01 ± 4.40	−5.33 ± 3.29	p = 0.108	p = 0.168	p = 0.885
IF	−5.28 ± 3.63	−6.86 ± 4.31			
EF	−5.14 ± 5.01	−6.49 ± 4.78			

$p < 0.05$*

Lima et al., 2020). This adjustment reflects a conservative strategy by individuals with FAI during landing. Furthermore, it facilitates effective force dispersion and absorption by increasing knee flexion (Blackburn and Padua, 2008). It is noteworthy that the increase in the varus angle is not mutually exclusive with the conservative drop landing strategy; rather, they complement each other, jointly promoting the optimization of the lower limb's biomechanical response. Moreover, this adjustment of the varus angle may help individuals with FAI maintain balance under unstable conditions, serving as an effective means of dispersing influence forces. Tamura et al. (2017) suggest that an appropriate varus angle at the knee during unilateral drop landing tasks can better absorb and cushion the shock, underscoring the importance of this strategy for managing joint stress. From the perspective of biomechanical adjustment theory that the rigid control associated with IF may increase joint stress, while the flexible control under EF conditions can improve impact dispersion, helping to reduce peak ground reaction forces and thereby protect the joints (Wang et al., 2023). Therefore, this finding from the study emphasizes the need to pay special attention to the stability and control ability of the unstable lower limb in the rehabilitation training of individuals with FAI.

Additionally, this study found that individuals with FAI exhibited a greater varus angle at the ankle joint on the unstable side during drop landing, potentially linking to the impaired stability of the ankle joint (Simpson et al., 2019). At the peak vGRF, the plantarflexion angle of the ankle joint under EF conditions was smaller, indicating that the EF strategy might contribute to a more stable ankle joint position during landing, which could indirectly lower the risk of injury. Furthermore, from a neuromuscular control perspective, the EF strategy may promote a more focused landing approach by directing attention to the external environment, potentially enhancing stability. rather than their body movements (Mulla and Keir, 2023), reducing reliance on the compromised stability of the ankle joint and promoting more natural whole-body coordination, which is crucial for optimizing motor control and reducing injury risk (Singh et al., 2022). The EF strategy, by directing the attention of individuals with FAI to the external environment rather than their body movements, may encourage them to adopt more stable and effective movement patterns. This is consistent with the findings of Bæktoft van Weert et al. (2023), which suggest that the EF strategy can lead to improved jumping and drop landing techniques. Similarly, Widenhoefer et al. (2019) argue that training with an emphasis on an EF fosters adaptation of the body's drop landing mechanisms. This strategy not only helps to optimize the movement control of the ankle joint but may also enhance the overall stability and coordination of the lower limbs. Therefore, incorporating the EF strategy in the design of rehabilitation training programs is of significant importance for enhancing the softness and stability of landing in individuals with

TABLE 3 Knee joint angles at IC and peak vGRF moments during single drop landing.

Variables	Unstable side	Stable side	Main effect		Interaction effect
			Focus	Side	Focus × side
IC					
Knee flexion (+)/extension (−) angle (°)					
Baseline	10.18 ± 4.11	8.39 ± 3.47	p = 0.156	p = 0.881	p = 0.061
IF	11.68 ± 4.56	12.64 ± 4.85			
EF	11.58 ± 5.53	12.76 ± 5.19			
Knee varus (+)/valgus (−) angle (°)					
Baseline	0.85 ± 2.38	0.57 ± 3.04	p = 0.275	p = 0.322	p = 0.890
IF	1.85 ± 2.68	1.39 ± 3.07			
EF	2.30 ± 2.93	1.55 ± 3.06			
Peak vGRF					
Knee flexion (+)/extension (−) angle (°)					
Baseline	28.29 ± 6.61	27.16 ± 7.40	**p = 0.012***	p = 0.746	p = 0.550
IF	28.67 ± 6.37	30.25 ± 6.64			
EF	36.94 ± 10.99	37.90 ± 11.86			
Knee varus (+)/valgus (−) angle (°)					
Baseline	1.44 ± 4.91	−1.46 ± 4.82	p = 0.778	**p = 0.028***	p = 0.549
IF	2.89 ± 5.67	−1.92 ± 4.87			
EF	2.63 ± 7.38	−2.95 ± 6.61			

$p < 0.05$*

FAI. By adjusting the focus of attention, the EF strategy may assist patients in achieving softer and more stable landings, reducing excessive motion and lowering the risk of injury.

The overall findings suggest that attentional focus strategies exert a significant influence on the biomechanics of the lower limbs among individuals with FAI, notably by promoting softer and more stable landings. The study, however, did not identify any interactive effects, implying that the impact of these strategies on both unstable and stable lower limb joints is consistent, which may indicate that individuals with FAI may benefit from the adjustment of attentional focus strategies during drop landing movements, regardless of whether it is the unstable or stable side.

4.2 Optimization of drop landing stability in individuals with FAI through attentional focus strategies

This study observed significant main effects of focus on both peak vGRF and the time to peak vGRF. Although individuals with FAI under baseline and IF conditions exhibited higher peak vGRF compared to the EF strategy, the time to reach peak vGRF was significantly reduced under the IF condition. Moreover, under the IF condition, individuals with FAI demonstrated greater k_{leg}, with a significant increase on the unstable side in terms of k_{leg}, indicating that the IF strategy may aid in enhancing the stability and support capacity of the lower limbs, reducing unstable elements during the drop landing process, thereby diminishing the injury risk for individuals with FAI during physical activity. However, under the EF strategy, individuals with FAI exhibited a longer duration to reach peak vGRF, attributable to the significant increase in knee flexion angle under the EF strategy. As the knee flexion angle increases, the influence force during drop landing is more effectively dispersed and absorbed, thereby extending the contact time with the ground (Harry et al., 2019). This extended contact time allows patients to adjust the posture of their lower limbs by increasing knee flexion, enabling a more effective cushioning against the influence from the ground, thereby augmenting the time to reach the peak vGRF (Almonroeder et al., 2020). The EF strategy, by prompting individuals with FAI to focus on the external environment rather than bodily sensations (Chua et al., 2021), may assist individuals with FAI in controlling the drop landing motion more gracefully, achieving a smoother force transmission process. Under this strategy, increasing the knee flexion angle not only aids in shock absorption but also provides patients with more time to adjust the posture of their lower limbs, refine the drop landing motion, thereby reducing the instantaneous influence on the lower limb joints. Therefore, the extended peak vGRF time under the EF strategy reflects a more cautious and controlled drop landing approach, which is a safer and more effective sports strategy for

TABLE 4 Ankle joint angles at IC and peak vGRF moments during single drop landing.

Variables	Unstable side	Stable side	Main effect		Interaction effect
			Focus	Side	Focus × side
IC					
Ankle dorsiflexion (+)/plantarflexion (–) angle (°)					
Baseline	−18.35 ± 5.85	−18.58 ± 5.65	p = 0.378	p = 0.069	p = 0.082
IF	−18.64 ± 6.25	−15.60 ± 6.06			
EF	−21.47 ± 5.53	−17.75 ± 5.09			
Ankle inversion (+)/eversion (–) angle (°)					
Baseline	5.66 ± 3.82	2.80 ± 4.24	p = 0.312	**p = 0.002***	p = 0.306
IF	7.42 ± 5.38	3.06 ± 3.53			
EF	8.28 ± 4.86	3.39 ± 3.31			
Peak vGRF					
Ankle dorsiflexion (+)/plantarflexion (–) angle (°)					
Baseline	11.15 ± 4.65	12.38 ± 6.11	**p = 0.017***	p = 0.376	p = 0.982
IF	7.65 ± 5.35	8.38 ± 3.63			
EF	13.33 ± 4.52	14.22 ± 8.02			
Ankle inversion (+)/eversion (–) angle (°)					
Baseline	0.13 ± 3.39	−0.27 ± 5.25	p = 0.521	p = 0.823	p = 0.628
IF	0.84 ± 4.37	0.16 ± 3.45			
EF	−0.69 ± 4.72	−0.25 ± 3.45			

p < 0.05*

individuals with FAI, consistent with the aforementioned discussion.

Attentional focus strategies may improve landing techniques in individuals with FAI by optimizing biomechanical responses related to softness and stability (Aiken and Becker, 2023). In particular, the EF strategy can better optimize drop landing stability (Vaz et al., 2019), and by reducing excessive activity of the ankle joint, may help to enhance the stability of the ankle joint and reduce the risk of sprains. Furthermore, the IF strategy, by prompting individuals with FAI to focus on the execution of drop landing movements, may help to reduce the influence force experienced by the joints and enhance the stability of movement. However, it should be noted that although the peak vGRF is higher under the IF condition, this does not necessarily imply an increased risk of injury; movements under this condition may require more attention to be allocated to internal bodily sensations, which could focus attention on movement execution and thereby enhance control and stability (Chen et al., 2023). Moreover, the time for individuals with FAI to reach peak vGRF is shortened under the IF condition, which may help to reduce unstable factors during the drop landing process, thereby enhancing stability. Additionally, increased kleg may help to provide better support and stability, further reducing the risk of injury. This finding is inconsistent with Wulf's traditional constraint-led action hypothesis (Wulf et al., 2001a; Wulf et al., 2001b), which suggests that IF would constrain the motor system by interfering with the automation of movement regulation, while EF might allow the motor system to self-organize more naturally, without interference from conscious control, leading to more effective performance and learning. However, according to the results of this study, both IF and EF strategies have produced positive effects on the movement performance of individuals with FAI. Therefore, this hypothesis may not be applicable to individuals with FAI, possibly because the ankle joint injury in this population leads to a lack of lower limb stability and support (Yin et al., 2016), making them more focused on the execution of movements when adopting the IF strategy, thereby improving movement precision and control.

Based on the aforementioned perspectives, the results of this study suggest that the EF strategy may encourage individuals with FAI to adopt a more conservative drop landing strategy, achieving a "soft landing" by increasing the knee flexion angle, thereby reducing direct influence and pressure on the knee joint. Furthermore, the EF strategy may also help to reduce excessive activity of the ankle joint and enhance its stability. At the same time, the IF strategy may help individuals with FAI to focus more on the execution of their movements upon drop landing, thereby improving the stability and support capacity of the lower limbs. These findings support Hypothesis 2 of this study, that appropriate attentional focus strategies have a positive effect on improving the biomechanical characteristics and motor control of individuals with FAI. However, both IF and EF strategies have certain drawbacks; while the IF

TABLE 5 Peak GRF and kleg during single drop landing.

Variables	Unstable side	Stable side	Main effect		Interaction effect
			Focus	Side	Focus × side
Peak vGRF (BW)					
Baseline	2.83 ± 0.64	2.42 ± 0.66	**$p < 0.001$***	$p = 0.078$	$p = 0.400$
IF	2.99 ± 1.17	2.77 ± 0.86			
EF	1.73 ± 0.42	1.70 ± 0.47			
Peak mGRF (BW)					
Baseline	0.13 ± 0.02	0.12 ± 0.01	$p = 0.445$	$p = 0.138$	$p = 0.306$
IF	0.13 ± 0.04	0.13 ± 0.04			
EF	0.14 ± 0.02	0.11 ± 0.04			
Peak lGRF (BW)					
Baseline	−0.14 ± 0.02	−0.13 ± 0.02	$p = 0.174$	$p = 0.821$	$p = 0.634$
IF	−0.14 ± 0.04	−0.14 ± 0.05			
EF	−0.12 ± 0.03	−0.13 ± 0.03			
Time to Peak vGRF(s)					
Baseline	0.07 ± 0.01	0.07 ± 0.01	**$p < 0.001$***	$p = 1.000$	$p = 0.691$
IF	0.06 ± 0.01	0.06 ± 0.02			
EF	0.09 ± 0.03	0.09 ± 0.03			
k_{leg} (BW/m)					
Baseline	19.79 ± 10.25	16.44 ± 9.83	**$p = 0.016$***	**$p = 0.041$***	$p = 0.506$
IF	16.58 ± 9.13	14.48 ± 7.40			
EF	10.08 ± 5.05	9.44 ± 4.23			

$p < 0.05$*

strategy may enhance stability, in some cases, excessive IF might increase muscle tension, thereby affecting the fluidity and naturalness of movement. The EF strategy could direct attention away from the execution details, potentially diminishing the fluidity of movements and impacting overall stability. Therefore, in rehabilitation training, the combined use of IF and EF strategies can enhance the adaptability and flexibility of patients. For instance, employing the IF strategy to establish correct movement patterns in individuals with FAI; utilizing the EF strategy to enhance their adaptability to the external environment. Furthermore, a tailored approach to selecting or integrating IF and EF strategies, customized to the unique conditions and rehabilitative aspirations of each individual with FAI, may represent the most efficacious strategy.

4.3 Clinical recommendations

This study elucidates the importance of employing diverse attentional focus strategies within personalized and comprehensive rehabilitation programs for individuals with FAI. When devising personalized rehabilitation plans for individuals with FAI, it is recommended that the IF strategy be employed for those who require the establishment of proper movement patterns and enhancement of lower limb stability; For those with FAI who need to optimize their adaptability to the external environment and the fluidity of movement, the EF strategy is recommended. By integrating IF and EF strategies, an efficacious rehabilitation training regimen can be crafted for individuals with FAI, enhancing movement execution, reducing injury risk.

4.4 Limitations

In this study, significant main effects of attention focus on hip joint flexion angle were observed at the IC and at the peak vGRF. Specifically, compared with the baseline, both IF and EF conditions demonstrated a trend of change in hip joint flexion angle. However, after Bonferroni correction, these changes did not reach statistical significance. This may imply that although different attention focus strategies have an influence on the hip joint flexion angle, the influence is not statistically robust and may be interfered with by sample characteristics, test conditions, or other uncontrolled variables.

5 Conclusion

The tailored application of IF and EF strategies exerted distinct influences on biomechanical outcomes in individuals with FAI. The IF, by directing attention to body movements, enhanced lower limb stability and support capabilities, which is crucial for reducing landing influence and improving joint shock absorption. Conversely, the EF, which diverts attention away from body sensations, encouraged a more conservative landing strategy characterized by increased knee flexion angles. This approach not only facilitated a softer landing by effectively dispersing impact forces over a longer contact time but also helped in minimizing the instantaneous stress on the lower limb joints. Collectively, integrating these strategies into FAI rehabilitation programs can optimize lower limb biomechanics and reduce the risk of reinjury.

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 humans were approved by Soochow university ethics committee. The studies were conducted in accordance with the local legislation and institutional requirements. The participants provided their written informed consent to participate in this study.

Author contributions

ZW: Conceptualization, Data curation, Formal Analysis, Funding acquisition, Investigation, Methodology, Project administration, Resources, Software, Supervision, Validation, Visualization, Writing–original draft, Writing–review and editing. LM: Data curation, Funding acquisition, Investigation, Writing–review and editing. ML: Data curation, Formal Analysis, Methodology, Writing–review and editing. LK: Data curation, Formal Analysis, Software, Writing–review and editing. JX: Data curation, Formal Analysis, Writing–review and editing. ZZ: Formal Analysis, Methodology, Writing–review and editing. XM: Formal Analysis, Writing–review and editing. QZ: Conceptualization, Data curation, Formal Analysis, Funding acquisition, Investigation, Methodology, Project administration, Resources, Software, Supervision, Validation, Visualization, Writing–original draft, Writing–review and editing.

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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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Does chronic ankle instability patients lead to changes in biomechanical parameters associated with anterior cruciate ligament injury during landing? A systematic review and meta-analysis

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Objective: This study aimed to determine if patients with chronic ankle instability (CAI) exhibit biomechanical changes associated with the increased risk of anterior cruciate ligament (ACL) injury during landing tasks.

Study Design: This study was conducted through systematic review and meta-analysis.

Data Sources: Searches were conducted in May 2024 across five electronic databases, including Web of Science, Scopus, PubMed, SPORTDiscus, and Cochrane Library.

Eligibility Criteria: Studies were included if they (1) involved subjects with CAI and healthy controls and (2) assessed biomechanical variables such as ground reaction forces, joint angles, and joint torques.

Results: Of the 675 identified studies, 171 were included in the review, and 13 were eligible for meta-analysis. The reviewed studies clearly defined research objectives, study populations, consistent participant recruitment, and exposures, and they used valid and reliable measures for outcomes. However, areas such as sample size calculation, study sample justification, blinding in assessments, and addressing confounders were not robust. This meta-analysis involved 542 participants (healthy group: n = 251; CAI group: n = 291). Compared with healthy individuals, patients with CAI exhibited a greater peak vertical ground reaction force (peak VGRF; SMD = 0.30, 95% CI: 0.07–0.53, $p = 0.009$), reduced hip flexion angles (SMD = −0.30, 95% CI: −0.51 to −0.17, $p < 0.0001$), increased trunk lateral flexion (SMD = 0.47, 95% CI: 0.05 to 0.9, $p = 0.03$), greater hip extension moments (SMD = 0.47, 95% CI: 0.09–0.84, $p = 0.02$), and increased knee extension moments (SMD = 0.39, 95% CI: 0.02–0.77, $p = 0.04$).

Conclusion: During landing tasks, patients with CAI demonstrate increased hip extension moments and knee extension moments, decreased hip flexion angles, increased peak VGRF, and increased trunk lateral flexion angles. These biomechanical variables are associated with an elevated risk of ACL injuries.

Systematic Review Registration: Identifier CRD42024529349.

KEYWORDS

chronic ankle instability, proximal, joint biomechanics, landing, anterior cruciate ligament

1 Introduction

Anterior cruciate ligament (ACL) injuries are among the most severe injuries in sports, which remarkably affect athletic performance (Davey et al., 2019). The incidence and cost of recovery from ACL injuries are high, with approximately 100,000 to 2,50,000 cases occurring annually in the United States alone (Hewett et al., 1999; Mancino et al., 2023). The estimated cost per injury is about $17,000, with an annual cost of approximately $646 million for surgeries and rehabilitation among females with ACL injuries (Hewett et al., 1999). Moreover, pain, functional limitations, and radiographic signs of osteoarthritis in the knee are evident 12–20 years post-injury (Lohmander et al., 2004; Myklebust and Bahr, 2005). The current paradigms for ACL injury mechanisms during landing include four main theories: Ligament dominance, Quadriceps dominance, Trunk dominance, and Leg dominance (Hewett et al., 2010). Based on these theories, previous scholars have identified eight biomechanical variables associated with ACL injury risk during landing (dynamics and kinematics in the horizontal, frontal, and sagittal planes, impact loads, and trunk movements) (Chou et al., 2023; Hewett et al., 2010). For instance, small knee and hip flexion angles in sagittal plane kinematics have been previously reported to correlate with a great risk of ACL injuries (Devita and Skelly, 1992; Lin et al., 2012; Padua et al., 2009; Trigsted et al., 2017) (Table 1). Identifying the biomechanical changes and influencing factors of individual ACL injury patterns is crucial for assessing functional capabilities and predicting subsequent injury risks post-return to sports.

Existing research suggests that chronic ankle instability (CAI) may be important predisposing factor for ACL injuries. Epidemiological surveys indicate that 52%–60% of patients with ACL injuries have a history of ankle sprains (Kramer et al., 2007; Söderman et al., 2001), and most ankle sprains develop into CAI (Theisen and Day, 2019). Studies by Theisen and Day also suggest that the altered lower limb biomechanics associated with CAI may increase the susceptibility to non-contact ACL injuries (Theisen and Day, 2019). The etiology of CAI is often attributed to injuries of the lateral ankle ligaments, primarily the anterior talofibular ligament, resulting from sprains, with or without damage to the calcaneofibular ligament (van Putte-Katier et al., 2015). In addition, CAI is a common sequela of recurrent ankle sprains (Gribble et al., 2016). About 73% of individuals with ankle sprains develop CAI (Theisen and Day, 2019). Patients with CAI often experience persistent symptoms of pain or weakness in the ankle (Gribble et al., 2013), which affects their quality of life and participation in physical activities (Arnold et al., 2011; Hubbard-Turner and Turner, 2015), as well as increases the risk of recurrent sprains and post-traumatic ankle osteoarthritis (Gribble et al., 2016; Moisan et al., 2017).

In recent years, CAI has received attention as a risk factor for ACL injuries during landing motions (Kikumoto et al., 2024). Current literature proposes that the mechanism of ACL injuries in patients with CAI is due to distal responses in the kinematic chain triggered by the ankle during landing tasks, which result in alterations in the proximal movement chain (knee joint, hip joint, and trunk) (Hertel and Corbett, 2019; Sueki et al., 2013). Munn et al. have shown that several somatosensory domains are impaired in patients with CAI, possibly due to ligament and joint proprioceptor damage during injury and potential neural damage post-ligament injury (Munn et al., 2010). Impairments in proprioceptors can lead to altered feedforward motor control

TABLE 1 Grouping of biomechanical variables.

Sagittal Plane Kinematics	Hip flexion angle↓(Devita and Skelly, 1992; Trigsted et al., 2017)	Knee flexion angle↓(Devita and Skelly, 1992; Trigsted et al., 2017)	Ankle dorsiflexion angle↓(Boden et al., 2009; Padua et al., 2009)
Sagittal Plane Kinetics	Hip extension moment↑(Devita and Skelly, 1992; Trigsted et al., 2017)	Knee extension moment↑(Devita and Skelly, 1992; Trigsted et al., 2017)	
Frontal plane kinematics	Hip abduction angle↓(Hewett et al., 2005)	Knee abduction angle↑(Hewett et al., 2005)	
Frontal plane Kinetics	knee adduction moment↑(Lin et al., 2012)		
Horizontal plane kinematics	Hip Internal rotation angle↑(Devita and Skelly, 1992; Trigsted et al., 2017)	Knee Internal rotation angle↑(Devita and Skelly, 1992; Trigsted et al., 2017)	
Horizontal plane Kinetics	Hip Internal rotation Moment↑(Devita and Skelly, 1992; Trigsted et al., 2017)	Knee Internal rotation Moment↑(Devita and Skelly, 1992; Trigsted et al., 2017)	
Impact loading	Peak VGRF↑(Devita and Skelly, 1992)	Loading rate↑(Paterno et al., 2007)	
Trunk mechanism	Trunk flexion↓(Paterno et al., 2007)	Trunk lateral flexion↑(Kristianslund et al., 2012)	

TABLE 2 Search strategy for each database.

Database	Web of Science	Scopus	PubMed	SPORTDiscus	Cochrane Library
Applied database fields used During The search	Topic (Title, abstract, author, keywords, and Keywords Plus)	Title Abstract, keyword	Title, Abstract	Title, Abstract	Title Abstract, keyword
Restrictions for the search	None				
Examples of the strategy Web of Science	(((TS=(chronic ankle instability OR ankle instability OR functional ankle instability OR mechanical ankle instability OR CAI OR FAI OR MAI)) AND TS=(lower limb OR lower extremity OR hip OR knee OR ankle OR trunk)) AND TS=(kinematic OR kinetics OR biomechanics OR Moment OR Torque OR dynamic OR angles OR moments OR forces OR ground reaction force OR GRF OR displacement)) AND TS=(stop jump OR stop-jump OR stop jumping OR stop-jumping OR 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)				
Examples of the strategy Scopus	(TITLE-ABS-KEY ("chronic ankle instability" OR "ankle instability" OR "functional ankle instability" OR "mechanical ankle instability" OR "CAI" OR "FAI" OR "MAI") AND TITLE-ABS-KEY ("lower limb" OR "lower extremity" OR "hip" OR "knee" OR "ankle" OR "trunk") AND TITLE-ABS-KEY ("kinematic" OR "kinetics" OR "biomechanics" OR "Moment" OR "Torque" OR "dynamic" OR "angles" OR "moments" OR "forces" OR "ground reaction force" OR "GRF" OR "displacement") AND TITLE-ABS-KEY ("stop jump" OR "stop-jump" OR "stop jumping" OR "stop-jumping" OR "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"))				
Examples of the strategy PubMed	(((chronic ankle instability [Title/Abstract] OR ankle instability [Title/Abstract] OR functional ankle instability [Title/Abstract] OR mechanical ankle instability [Title/Abstract] OR CAI [Title/Abstract] OR FAI [Title/Abstract] OR MAI [Title/Abstract]) AND (lower limb [Title/Abstract] OR lower extremity [Title/Abstract] OR hip [Title/Abstract] OR knee [Title/Abstract] OR ankle [Title/Abstract] OR trunk [Title/Abstract])) AND (kinematic [Title/Abstract] OR kinetics [Title/Abstract] OR biomechanics [Title/Abstract] OR Moment [Title/Abstract] OR Torque [Title/Abstract] OR dynamic [Title/Abstract] OR angles [Title/Abstract] OR moments [Title/Abstract] OR forces [Title/Abstract] OR ground reaction force [Title/Abstract] OR GRF [Title/Abstract] OR displacement [Title/Abstract])) AND (stop jump [Title/Abstract] OR stop-jump [Title/Abstract] OR stop jumping [Title/Abstract] OR stop-jumping [Title/Abstract] OR land [Title/Abstract] OR landing [Title/Abstract] OR jump land [Title/Abstract] OR jump landing [Title/Abstract] OR jump-land [Title/Abstract] OR jump-landing [Title/Abstract] OR drop-vertical jump [Title/Abstract] OR single-leg landing [Title/Abstract] OR single-leg land [Title/Abstract] OR jump [Title/Abstract] OR jumping [Title/Abstract])				
Examples of the strategy SPORTDiscus	TI (chronic ankle instability OR ankle instability OR functional ankle instability OR mechanical ankle instability OR CAI OR FAI OR MAI) AND TI (lower limb OR lower extremity OR hip OR knee OR ankle OR trunk) AND TI (kinematic OR kinetics OR biomechanics OR Moment OR Torque OR dynamic OR angles OR moments OR forces OR ground reaction force OR GRF OR displacement) AND TI (stop jump OR stop-jump OR stop jumping OR stop-jumping OR 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)				
	AB (chronic ankle instability OR ankle instability OR functional ankle instability OR mechanical ankle instability OR CAI OR FAI OR MAI) AND AB (lower limb OR lower extremity OR hip OR knee OR ankle OR trunk) AND AB (kinematic OR kinetics OR biomechanics OR Moment OR Torque OR dynamic) AND AB (stop jump OR stop-jump OR stop jumping OR stop-jumping OR 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)				
Examples of the strategy Cochrane Library	chronic ankle instability OR ankle instability OR functional ankle instability OR mechanical ankle instability OR CAI OR FAI OR MAI in Title Abstract Keyword AND lower limb OR lower extremity OR hip OR knee OR ankle OR trunk in Title Abstract Keyword AND kinematic OR kinetics OR biomechanics OR Moment OR Torque OR dynamic OR angles OR moments OR forces OR ground reaction force OR GRF OR displacement in Title Abstract Keyword AND stop jump OR stop-jump OR stop jumping OR stop-jumping OR 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 in Title Abstract Keyword - (Word variations have been searched)				

mediated by spinal or supraspinal mechanisms, affecting centrally mediated motor control strategies (Brown et al., 2004; Caulfield and Garrett, 2002; Hass et al., 2010). These outcomes may lead to changes in the proximal movement chain to compensate for ankle instability and functional impairment (Bullock-Saxton, 1994; Koshino et al., 2013). Such compensatory phenomena in proximal joints could cause ACL injuries (Jeon et al., 2021; Xu et al., 2022). For example, Terada et al. found reduced knee flexion at the peak of the anterior tibial shear force in patients with CAI (Terada et al., 2014), and Jeon et al.'s meta-analysis identified greater vertical ground reaction forces (VGRF) upon landing in patients with CAI (Jeon et al., 2021), both results were associated with ACL injury risk. However, the aforementioned studies identified only a single biomechanical variable difference associated with ACL injuries, probably due to the limited outcome indicators set in previous research, which failed to report more biomechanical differences related to ACL injuries. Consequently, many studies based on similar measurements have produced inconsistent results, and the adoption of a single variable alone is insufficient to objectively describe the ACL injury mechanisms in CAI patients. The lack of combined analyses of variables related to ACL injuries, which limits the effective examination of how proximal adaptations caused by CAI lead to ACL injuries.

A 13-year epidemiological study revealed that landing maneuvers are a common cause of ACL injuries in sports with frequent jumping activities, such as basketball and volleyball (Agel et al., 2005). Krosshaug et al. (2006) analyzed videos of 39 ACL injuries during basketball games and found that most injuries occurred during non-contact landing maneuvers. The landing action, given its high impact and frequent occurrence, warrants particular attention (Kim et al., 2017; Konradsen and Voigt, 2002). Moreover, landing provides a controlled and easily standardizable experimental condition, facilitating accurate assessment and analysis of biomechanical parameters. Considering the commonality, representativeness, high impact, and robustness of landing measurements in ACL injury studies, this research focuses on the landing maneuvers of patients with CAI.

Therefore, a systematic review and meta-analysis were conducted to investigate whether biomechanical changes related to ACL injury risk are present in patients with CAI during landing tasks. Previous studies have found that patients with CAI often fail to effectively utilize the lower limb buffering mechanisms during landing owing to ankle instability (Jeon et al., 2021). Additionally, changes have been observed in the proximal movement chain of the lower limbs in CAI patients (Bullock-Saxton, 1994; Koshino et al., 2013). We hypothesized that compared with healthy individuals, those with CAI will exhibit a greater peak VGRF and an upright body posture (i.e., reduced knee and hip flexion), leading to an increased risk of ACL injury.

2 Methods

This review was conducted in accordance with the PRISMA guidelines for systematic reviews and meta-analyses. It was prospectively registered in the international database for systematic reviews. (PROSPERO registration number: CRD42024529349).

2.1 Identification of ACL injury-related variables

Four theories based on the mechanism of landing ACL injury (Ligament dominance; Quadriceps dominance; Trunk dominance; leg dominance) (Hewett et al., 2010), and previous prospective and case-control studies identified eight biomechanical variables associated with ACL injury risk (Chou et al., 2023; Hewett et al., 2010). These variables have been adopted in several high-quality studies (Chou et al., 2023), thereby proving their reliability. Table 1 illustrates the detrimental directionality of biomechanical variables in each construct (high injury risk). Following data extraction, clarifying which outcome indicators are used for correlation analysis is crucial, serving as the bridge to determine whether CAI is associated with ACL injuries.

2.2 Data sources and search strategy

In May 2024, a systematic search was conducted across five electronic databases: Web of Science, Scopus, PubMed, SPORTDiscus, and Cochrane Library. Keywords related to CAI and biomechanical variables were used to identify relevant articles. Keywords for each category were combined using the Boolean operator "OR" and then combined across categories using "AND" for each database search. There was no restriction on publication date. Boolean logic was utilized for all database searches: (chronic ankle instability OR ankle instability OR functional ankle instability OR mechanical ankle instability OR CAI OR FAI OR MAI) AND (lower limb OR lower extremity OR hip OR knee OR ankle OR trunk) AND (kinematic OR kinetics OR biomechanics OR Moment OR Torque OR dynamic) AND [1) stop jump: (stop jump OR stop-jump OR stop jumping OR stop-jumping), 2) 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)]. The detailed search strategy is presented in Table 2.

2.3 Inclusion criteria

The search results were independently screened by two reviewers (Z.H., B.Y.) on the basis of the predetermined inclusion and exclusion criteria, with disputes resolved by the corresponding author (H.Z.) if necessary. Studies were included if they met the following criteria: In accordance with the PICOS guidelines, studies meeting the following criteria were included: P: The study population consists of individuals with CAI and healthy control subjects. I: Participants perform landing tasks as an intervention measure. C: CAI patients are compared with a healthy control group. O: The reported outcomes are biomechanical variables, such as ground reaction forces, joint angles, and joint torques. S: Cross-sectional studies are included. Exclusion criteria included the following: 1) variables unrelated to ACL injury outcome indicators (Table 1); 2) task location is not on a flat surface (differences in biomechanics of landing on flat and sloped surfaces may affect the robustness of study results. Additionally, sloped landings are uncommon in real-life activities.); 3) studies employing one-dimensional statistical parametric mapping analysis (Data cannot be pooled for meta-analysis); 4) review articles, editorials, speeches, commentaries, abstracts, case studies, surgical procedures, and non-peer-reviewed articles.

2.4 Study selection

Titles and abstracts were screened by two independent reviewers, with any discrepancies resolved through consultation with a third author. If eligibility could not be determined from the title and abstract, then full texts were obtained and reviewed. Cohen's kappa coefficient (κ) and percentage agreement were used to assess inter-reviewer consistency. κ was interpreted on the basis of Landis and Koch's standards: Values less than 0 indicated no agreement, 0–0.20 indicated slight agreement; 0.21–0.40 indicated fair agreement; 0.41–0.60 indicated moderate agreement; 0.61–0.80 indicated substantial agreement; and 0.81–1 indicated almost perfect agreement (Landis and Koch, 1977). Full texts of all eligible studies were retrieved, and data such as demographics (e.g., gender and age), sample size, and biomechanical variable results (e.g., standardized ground reaction forces, joint angles, and standardized joint torques) were extracted.

2.5 Quality assessment and analysis

The quality of included studies was assessed using the NIH quality assessment tool for observational cohort and cross-sectional studies. Two researchers (Z.H. and B.Y.) independently evaluated the quality of each study, with any disagreements resolved through discussion until consensus was reached.

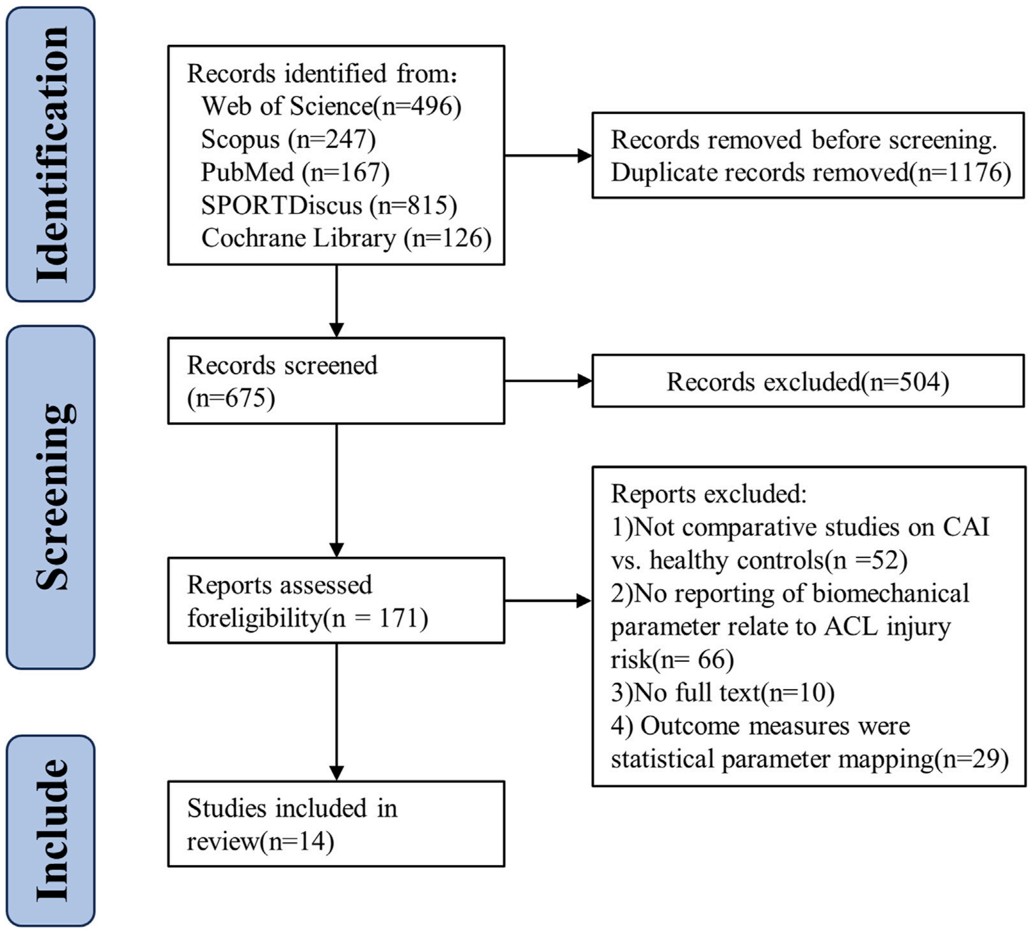

FIGURE 1
Flow chart of systematic reviews and meta-analyses.

2.6 Publication bias

Publication bias was assessed using funnel plots and Egger's regression test. Visual inspection of funnel plot symmetry indicated an unbiased range of effect sizes among the included studies. Asymmetry in the funnel plot suggested bias toward specific effect sizes or sample sizes. Egger's regression test quantified bias using the effect size of each included study, with a *p*-value less than 0.05 indicating significant publication bias (Borenstein et al., 2021).

2.7 Sensitivity analysis

Sensitivity analyses were conducted on studies included in the meta-analysis to determine if the exclusion of influential studies affected the overall effect results. Thus, the most influential studies were excluded from the meta-analysis and analysis was repeated.

2.8 Data extraction and statistical analysis

Data were extracted by two authors (Z.H., B.Y.) using standardized forms from each included study. The extracted data were as follows: 1) sample description, encompassing the sample size, age, and gender; 2) information related to CAI, including the definition of CAI and scores from the Cumberland Ankle Instability Tool; 3) biomechanical variables related to outcome indicators (Table 1); 4) standardized results, including means, SDs, *p*-values, and standardized difference in effect sizes; 5) task conditions, specifically limited to landing tasks performed on flat surfaces. The data extracted from all included studies were then synthesized for comprehensive analysis. Statistical analysis compared biomechanical variables between patients with CAI and healthy individuals as the effect sizes for meta-analysis. To avoid errors caused by multiple biomechanical variables within each structure influencing the final outcome, we conducted separate meta-analyses for biomechanical outcome indicators within each structure (e.g., separate meta-analyses for Trunk flexion and trunk lateral flexion within the trunk mechanism structure). Additionally, to prevent reliance on effect sizes from the same study, we averaged the effect sizes based on specific task conditions (e.g., averaging the parameters of a biomechanical variable at different stages during a landing task) (Borenstein et al., 2021).

The overall effect size for each variable was calculated using sample size, means, standard deviations, mean differences, and *p*-values. Meta-analyses were performed using Review Manager V.5.3 (Copenhagen, Denmark: The Nordic Cochrane Centre, The Cochrane Collaboration, 2014). Owing to variations in drop platform

TABLE 3 Outcome indicators determined.

Sagittal Plane Kinematics of the Lower Limb	Hip flexion angle↓(Devita and Skelly, 1992; Trigsted et al., 2017)	Knee flexion angle↓(Devita and Skelly, 1992; Trigsted et al., 2017)	Ankle dorsiflexion angle↓(Boden et al., 2009; Padua et al., 2009)
Sagittal Plane Kinetics of the Lower Limb	Hip extension moment↑(Devita and Skelly, 1992; Trigsted et al., 2017)	Knee extension moment↑(Devita and Skelly, 1992; Trigsted et al., 2017)	
Frontal plane kinematics of the Lower Limb	Hip abduction angle↓(Hewett et al., 2005)	Knee abduction angle↑(Hewett et al., 2005)	
Impact loading	Peak VGRF↑(Devita and Skelly, 1992)	Loading rate↑(Paterno et al., 2007)	
Trunk mechanism	Trunk flexion↓(Paterno et al., 2007)	Trunk lateral flexion↑(Kristianslund et al., 2012)	

heights and the differences between single and double foot landings among the included studies, ensuring complete consistency across multicenter and multiprotocol studies is challenging. Therefore, this study calculated the effect size of each biomechanical variable using the standardized mean difference (SMD) with a 95% confidence interval (CI). A random effect model was employed. Heterogeneity among the included studies was assessed using the I^2 statistic.

3 Results

3.1 Study selection and characteristics

An initial search across five databases yielded 1,615 articles. After duplicates were removed, 675 studies remained. Title and abstract screening by two reviewers eliminated 504 studies (consistency rate = 97.62%, κ = 0.49), and 171 studies were assessed for eligibility through full-text review. After this review, 157 studies were excluded, leaving 14 studies included in the meta-analysis (consistency rate = 100%, κ = 1). The detailed exclusion process at each stage is depicted in Figure 1. A total of 564 participants were involved in this study (healthy group: n = 262; CAI group: n = 302), with detailed characteristics of the included studies provided in Table 3. For variables related to ACL injury, seven studies were included on peak VGRF (Caulfield and Garrett, 2004; De Ridder et al., 2015; Jeon and Park, 2021; Lee et al., 2017; Watabe et al., 2022; Watanabe et al., 2021; Zhang et al., 2012), three studies on loading rate (De Ridder et al., 2015; Jeon and Park, 2021; Lee et al., 2017); nine and ten studies on hip flexion angle (Brown et al., 2011; Caulfield and Garrett, 2004; Gribble and Robinson, 2009; Han et al., 2023; Jeon and Park, 2021; Sagawa et al., 2024; Terada et al., 2015; Terada et al., 2014; Watabe et al., 2022; Watanabe et al., 2021) and knee flexion angle (Brown et al., 2011; Caulfield and Garrett, 2004; Gribble and Robinson, 2009; Han et al., 2023; Jeon and Park, 2021; Sagawa et al., 2024; Terada et al., 2015; Terada et al., 2014; Watabe et al., 2022; Watanabe et al., 2021), respectively. Ankle flexion angle is covered in 10 studies (Gribble and Robinson, 2009; Han et al., 2023; Jeon and Park, 2021; Kipp and Palmieri-Smith, 2012; Lee et al., 2017; Sagawa et al., 2024; Terada et al., 2015; Watabe et al., 2022; Watanabe et al., 2021; Zhang et al., 2012); and two studies each on hip extension moment (Jeon and Park, 2021; Watanabe et al., 2021) and knee extension moment (Jeon and Park, 2021; Watanabe et al., 2021). Four studies on hip abduction angle (Brown et al., 2011; Jeon and Park, 2021; Sagawa et al., 2024; Terada et al., 2015) and three studies on knee abduction angle (De Ridder et al., 2015; Jeon and Park, 2021; Lee et al., 2017); two studies on trunk flexion (Brown et al., 2011; Watabe et al., 2022) and two studies on trunk lateral flexion (Brown et al., 2011; Watabe et al., 2022). No other biomechanical variables related to ACL injuries were reviewed in this study (Table 4).

3.2 Quality assessment

The results of the quality assessment tool developed by NIH for observational cohorts and cross-sectional studies ranged from five to 9 (Table 5). The evaluation results highlight some apparent strengths and limitations of the included literature. The strengths lie in the studies having clear objectives, defined study populations, and consistent participant recruitment, and using valid and reliable methods for exposure definition and outcome measurement. These aspects are important indicators of high research quality, suggesting that the design and execution of the studies were successful in these respects. However, some shortcomings remain. Only two studies calculated the sample size and determined an appropriate study sample size, which could affect the statistical power of the results and the reliability of the conclusions. Furthermore, the lack of assessment of blinding implementation and insufficient justification for confounding variables could lead to bias and errors in the study outcomes.

3.3 Publication bias

Publication bias was detected in the trunk flexion angle (p = 0.007). No publication bias was found for the remaining variables (Supplementary Appendix S1).

3.4 Heterogeneity

Heterogeneity was present in the ankle dorsiflexion angle (Figure 2C), loading rate (Figure 3A), knee flexion angle (Figure 2B), and trunk flexion (Figure 4B).

3.5 Study findings

Our research findings indicated that patients with CAI exhibited a greater maximum (peak VGRF; SMD = 0.30, 95% CI: 0.07–0.53, p = 0.009), a reduced hip flexion angle (SMD = −0.34,95% CI:

TABLE 4 Study characteristics and participant demographics.

First Author (year)	Test method	Study type	Population (total/male/female)	Outcome measures
Watanabe et al. (2021)	single-leg landing (0.3 m,0.4 m,0.5 m)	cross-sectional	26 Competitive collegiate athletes; Control group = 13/7/4 Age = 20.6 ± 2.1 CAI group = 13/7/4 Age = 21.6 ± 1.6	*Kinematics* Peak knee Flexion angle (0.3 m,0.4 m,0.5 m) Peak hip flexion angle (0.3 m,0.4 m,0.5 m) *Kinetics* Peak knee extension moment (0.3 m,0.4 m,0.5 m) Peak hip extension moment (0.3 m,0.4 m,0.5 m) *Impact Loading* Peak VGRF (0.3 m,0.4 m,0.5 m)
Gribble and Robinson (2009)	double-leg take-off jump with a landing on a single limb	cross-sectional	Control group = 19/10/9 Age = 23.1 ± 3.9 CAI group = 19/10/9 Age = 20.3 ± 2.9	*Kinematics* Peak knee flexion angle (Injured side) Peak hip flexion (Injured side)
Zhang et al. (2012)	drop landing (0.6 m)	cross-sectional	Control group = 10Age = 24.1 (5.4) CAI group = 10 Age = 24.8 (5.7)	*Impact loading* 2nd peak vertical GRF
Terada et al. (2014)	vertical stop jump (50% of Vertmax)	case–control experiment design	Thirty-eight physically active participants Control group = 19/10/9 Age = 21.32 ± 4.04) CAI group = 19/10/9 Age = 20.11 ± 1.63	*Kinematics* Knee sagittal plane angle at Peak anterior tibial shear force (ATSF) Hip sagittal plane angle at Peak ATSF
Lee et al. (2017)	single-leg drop landing	Controlled laboratory	28 competitive taekwondo athletes Control group = 14/14 Age = 21.21 ± 2.08 CAI group = 14/14 Age = 20.07 ± 0.27	*Impact loading* Peak VGRF Loading rate
De Ridder et al. (2015)	Vertical drop (0.4 m)	cross-sectional	Control group = 30/12/18 Age = 25.7 ± 1.8 CAI group = 38/19/19 Age = 22.1 ± 3.4	*Impact Loading* Peak vertical GRF Loading rate
Caulfield and Garrett (2004)	single leg vertical drop (0.4 m)	cross-sectional	Control group = 10/10 Age = 22.6 ± 4.6 FAI group = 14/14 Age = 26.6 ± 6.3	*Impact loading* Peak vertical GRF
Jeon and Park (2021)	Drop landing (0.3 m)	cross-sectional	Control group = 18/18 Age = 20.22 ± 2.29 CAI group = 16/16 Age = 20.19 ± 1.47	*Kinematics* Hip flexion joint angle; hip abduction joint angle; Knee Flexion joint angle Knee Valgus joint angle *Kinetics* Hip flexion Joint moment Knee flexion Joint moment *Impact loading* Max vGRF Loading rate
Watabe et al. (2022)	proactive condition (single-leg landings) reactive condition (side-step cutting, 60° side-step cutting, single-leg landing, and forward stepping)	cross-sectional	28 physically active individuals; Control group = 14/Age = 21.5 ± 1.3 CAI group = 14/Age = 21.4 ± 1.4	*Kinematics* Maximum Right lateral trunk flexion in (PRO,REA) Maximum Trunk flexionin in (PRO,REA) Maximum Hip flexion in (PRO,REA) Maximum Knee flexion in (PRO,REA) Maximum Ankle dorsiflexion in (PRO,REA) *Impact loading* Vertical ground reaction force in (PRO,REA)
Brown et al. (2011)	single-leg landing in the anterior, lateral, and medial directions (50% of Vertmax)	cross-sectional	68 recreationally active participants Control group = 24/12/12male's Age = 19.8 ± 13/female's Age = 20.2 ± 1.0 MAI group = 21/8/13 male's Age = 18.6 ± 3.3/female's Age = 19.9 ± 1.0 FAI group = 23/11/12 male's Age = 20.5 ± 1.7/female's Age = 20.1 ± 1.5	*Kinematics* Knee flexion–extension angle Knee abduction–adduction angle Hip flexion angle Hip abduction–adduction angle Trunk flexion angle Trunk lateral flexion angle

(Continued on following page)

TABLE 4 (*Continued*) Study characteristics and participant demographics.

First Author (year)	Test method	Study type	Population (total/male/ female)	Outcome measures
Han et al. (2023)	single-leg drop-landing	cross-sectional	44 physically active individuals Control group = 22/11/11 Age = 23.4 ± 2.6) CAI group = 22/11/11 Age = 23.4 ± 2.4	*Kinematics* Knee flexion angle at initial contact in (Anticipated,Unanticipated) Maximum knee flexion angle in (Anticipated,Unanticipated) Knee displacement in (Anticipated,Unanticipated) Hip flexion angle at initial contact in (Anticipated,Unanticipated) Maximum hip flexion angle in (Anticipated,Unanticipated) Hip displacement in (Anticipated,Unanticipated)
Terada et al. (2015)	single-leg drop landing (0.3 m) conditions: (1) looking-down and (2) looking-up	Controlled laboratory	Thirty-eight physically active participants Control group = 19/6/13 Age = 20.58 ± 2.32 CAI group = 19/11/8 Age = 21.68 ± 4.82	*Kinematics* Hip sagittal plane angle at initial contact (looking-up,looking-down) Hip frontal plane at initial contact (looking-up,looking-down) Knee sagittal plane at initial contact (looking-up,looking-down) Knee frontal plane at initial contact (looking-up,looking-down)
Sagawa et al. (2024)	Single-leg Lateral Drop Landing (0.2 m)	cross-sectional	Control group = 19/19/0 Age = 23.9 ± 2.8 CAI group = 19/19/0 Age = 25.3 ± 2.9	*Kinematics* Hip Flexion maximum angle during 200 m interval post-landing Hip Abduction maximum angle during 200 m interval post-landing Knee Flexion maximum angle during 200 m interval post-landing

−0.53 to −0.19, $p < 0.0001$), (Figure 4A) and an increased trunk lateral flexion angle (SMD = 0.47,95% CI: 0.05–0.9, $p = 0.03$) during landing tasks. In Addition, these patients showed increased hip extension moments (SMD = 0.47,95% CI: 0.09–0.84, $p = 0.02$) and greater knee extension moments (SMD = 0.39,95% CI: 0.02–0.77, $p = 0.04$) (Figures 2C, D) compared with healthy controls. By contrast, no significant differences in the loading rate, knee flexion angle, trunk flexion angle, knee abduction angle (Figure 3B), ankle dorsiflexion angle (Figure 2E) and hip abduction angle were found between patients with CAI and healthy individuals. The sensitivity analysis results reveal that the outcomes for Knee extension moment and Loading rate are significantly influenced by individual studies (Supplementary Appendix S4). Specifically, removing the study by Jeon and Park (2021) led to a significant change in the p-value for Knee extension moment ($p < 0.05$), and removing Lee et al. (2017) similarly affected the p-value for Loading rate ($p < 0.05$). Due to the limited number of studies included, further subgroup analysis to explore these differences was not feasible. This limitation highlights the need for cautious interpretation of our conclusions. Additionally, the results for other variables remained consistent and unchanged in the sensitivity analysis.

4 Discussion

This systematic review aimed to determine whether changes in lower limb biomechanics during landing tasks in the CAI population increase the risk of ACL injury compared with that in a healthy population. This meta-analysis included data from 14 studies (Brown et al., 2011; Caulfield and Garrett, 2004; De Ridder et al., 2015; Gribble and Robinson, 2009; Han et al., 2023; Jeon and Park, 2021; Kipp and Palmieri-Smith, 2012; Lee et al., 2017; Sagawa et al., 2024; Terada et al., 2015; Terada et al., 2014; Watabe et al., 2022; Watanabe et al., 2021; Zhang et al., 2012), which compared the landing biomechanics between patients with CAI and healthy individuals. Our findings indicated that patients with CAI exhibited a greater peak VGRF during landing compared with healthy individuals, which was consistent with our initial hypothesis. A decreased hip flexion angle was also observed, partially supporting our hypothesis that the body was upright, although no significant difference was found in knee flexion angles. Additionally, an increased trunk lateral flexion, an increased hip extension moment, and an increased knee extension moment were observed in patients with CAI.

4.1 Sagittal plane kinematics and kinetics of the lower limb

Our study revealed that individuals with CAI landed with decreased hip flexion angles and increased knee and hip extension moments. However, contrary to our expectations, patients with CAI did not exhibit reduced knee flexion angles or dorsiflexion angles. This finding is also in contrast to the findings of Chan et al. (2022) and Aaron and James (Theisen and Day, 2019), who reported reduced knee flexion and ankle dorsiflexion in similar contexts (Chan et al., 2022). This inconsistency may stem from our meta-analysis incorporating anticipated and unanticipated landing conditions (Han et al., 2023; Watabe et al., 2022) and

TABLE 5 Methodological quality score using the NIH quality assessment tool of relevant studies.

First Author (year)	Q1	Q2	Q3	Q4	Q5	Q6	Q7	Q8	Q9	Q10	Q11	Q12	Q13	Q14	Total (Yes)
Watanabe et al. (2021)	Yes	Yes	CD	Yes	Yes	No	No	NA	Yes	No	Yes	No	NA	No	6
Kipp and Palmieri-Smith (2012)	Yes	Yes	CD	Yes	No	No	No	NA	Yes	No	Yes	No	NA	No	5
Gribble and Robinson (2009)	Yes	Yes	CD	Yes	No	No	No	NA	Yes	No	Yes	No	NA	No	5
Zhang et al. (2012)	Yes	Yes	CD	Yes	No	No	No	NA	Yes	No	Yes	No	NA	No	5
Terada et al. (2014)	Yes	Yes	CD	Yes	No	No	No	NA	Yes	No	Yes	No	NA	No	5
Lee et al. (2017)	Yes	Yes	CD	Yes	No	No	No	NA	Yes	No	Yes	No	NA	No	5
De Ridder et al. (2015)	Yes	Yes	CD	Yes	No	No	No	NA	Yes	No	Yes	No	NA	No	5
Caulfield and Garrett (2004)	Yes	Yes	CD	Yes	No	No	No	NA	Yes	No	Yes	No	NA	No	5
Jeon and Park (2021)	Yes	Yes	CD	Yes	No	No	No	NA	Yes	No	Yes	No	NA	No	5
Watabe et al. (2022)	Yes	Yes	CD	Yes	No	No	No	NA	Yes	No	Yes	No	NA	No	5
Han et al. (2023)	Yes	Yes	CD	Yes	No	No	No	NA	Yes	No	Yes	No	NA	No	5
Terada et al. (2015)	Yes	Yes	CD	Yes	No	No	No	NA	Yes	No	Yes	No	NA	No	5
Brown et al. (2011)	Yes	Yes	CD	Yes	No	No	No	NA	Yes	No	Yes	No	NA	No	5
Sagawa et al. (2024)	Yes	Yes	CD	Yes	Yes	No	No	NA	Yes	No	Yes	No	NA	No	6

Q1: was the research question or objective in this paper clearly stated?, Q2: was the study population clearly specified and defined?, Q3: was the participation rate of eligible persons at least 50%?, Q4: were all the subjects selected or recruited from the same or similar populations (including the same time period)? Were inclusion and exclusion criteria for being in the study prespecified and applied uniformly to all participants?, Q5: was a sample size justification, power description, or variance and effect estimates provided?, Q6: for the analyses in this paper, were the exposure(s) of interest measured prior to the outcome(s) being measured?, Q7: was the timeframe sufficient so that one could reasonably expect to see an association between exposure and outcome if it existed?, Q8: for exposures that can vary in amount or level, did the study examine different levels of the exposure as related to the outcome (e.g., categories of exposure, or exposure measured as continuous variable)?, Q9: were the exposure measures (independent variables) clearly defined, valid, reliable, and implemented consistently across all study participants?, Q10: was the exposure(s) assessed more than once over time?, Q11: were the outcome measures (dependent variables) clearly defined, valid, reliable, and implemented consistently across all study participants?, Q12: were the outcome assessors blinded to the exposure status of participants?, Q13: was loss to follow-up after baseline 20% or less?, Q14: were key potential confounding variables measured and adjusted statistically for their impact on the relationship between exposure(s) and outcome(s)? CD: cannot determine,NA: notapplicable.

changes in visual focus during landing (Terada et al., 2015). Analyzing multiple states together led to high heterogeneity, but this result was considered acceptable because it reflected the different dynamic states encountered during athletic activities. Moreover, our findings suggested that patients with CAI developed neuromuscular feedforward mechanisms that resulted in upright postures and increased hip extension moments during landing, thereby supporting our study outcomes (Figure 2) (Beckman and Buchanan, 1995; Kim et al., 2017).

High knee extension moments during landing are associated with increased risk of ACL injuries (Alentorn-Geli et al., 2009). This injury mechanism is prevalent among female athletes (Karita et al., 2017). The primary reason is that women typically have wider hips, which leads to a larger Q-angle (the angle between the lateral femoral axis and the tibial axis) compared with men (Mizuno et al., 2001). An increased Q-angle enhances the lateral pulling force of the quadriceps, resulting in greater knee extension torque (Heiderscheit et al., 2000). Additionally, women's weaker muscle strength and neuromuscular control may contribute to increased knee extension torque during landing (Hewett et al., 2005). Cadaver studies have shown that intense quadriceps contractions can cause ACL ruptures (DeMorat et al., 2004). Furthermore, during high-speed eccentric contractions and quadriceps activation during landing, the ACL is subjected to greater forces compared with those observed during maximum isometric contractions of the quadriceps (Griffin et al., 2000). Previous research also supported our findings, indicating that patients with CAI exhibited a higher quadriceps/hamstring co-activation ratio during inclined plane landings compared with healthy individuals (Li et al., 2017). During landing, the hamstring's force is diminished, and the quadriceps play a dominant role. Previous studies have shown that greater hamstring activation can reduce the ACL load exerted by the quadriceps and provide dynamic knee stability by resisting anterior tibial translation, lateral tibial translation, and transverse tibial rotation (Renström et al., 1986; Withrow et al., 2008). Therefore, the increased knee extension moment observed during landing in patients with CAI may be a potential mechanism leading to ACL injuries in this population.

Previous research has revealed that the contraction of the knee and hip extensors does not decrease VGRF during the landing contact phase; by contrast, the associated energy is transferred to the bones and ligaments, thereby increasing joint contact stress and the risk of ACL and meniscal injuries (Mills et al., 2009). In addition, this phenomenon has been observed in the biomechanical factors related to ACL injury and knee joint energy absorption. At the terminal phase of landing, knee joint energy absorption is inversely related to VGRF and hip extension moments and directly related to peak hip flexion angles (Norcross et al., 2010). These findings indicated that high hip extension moments and reduced hip flexion angles could lead to insufficient

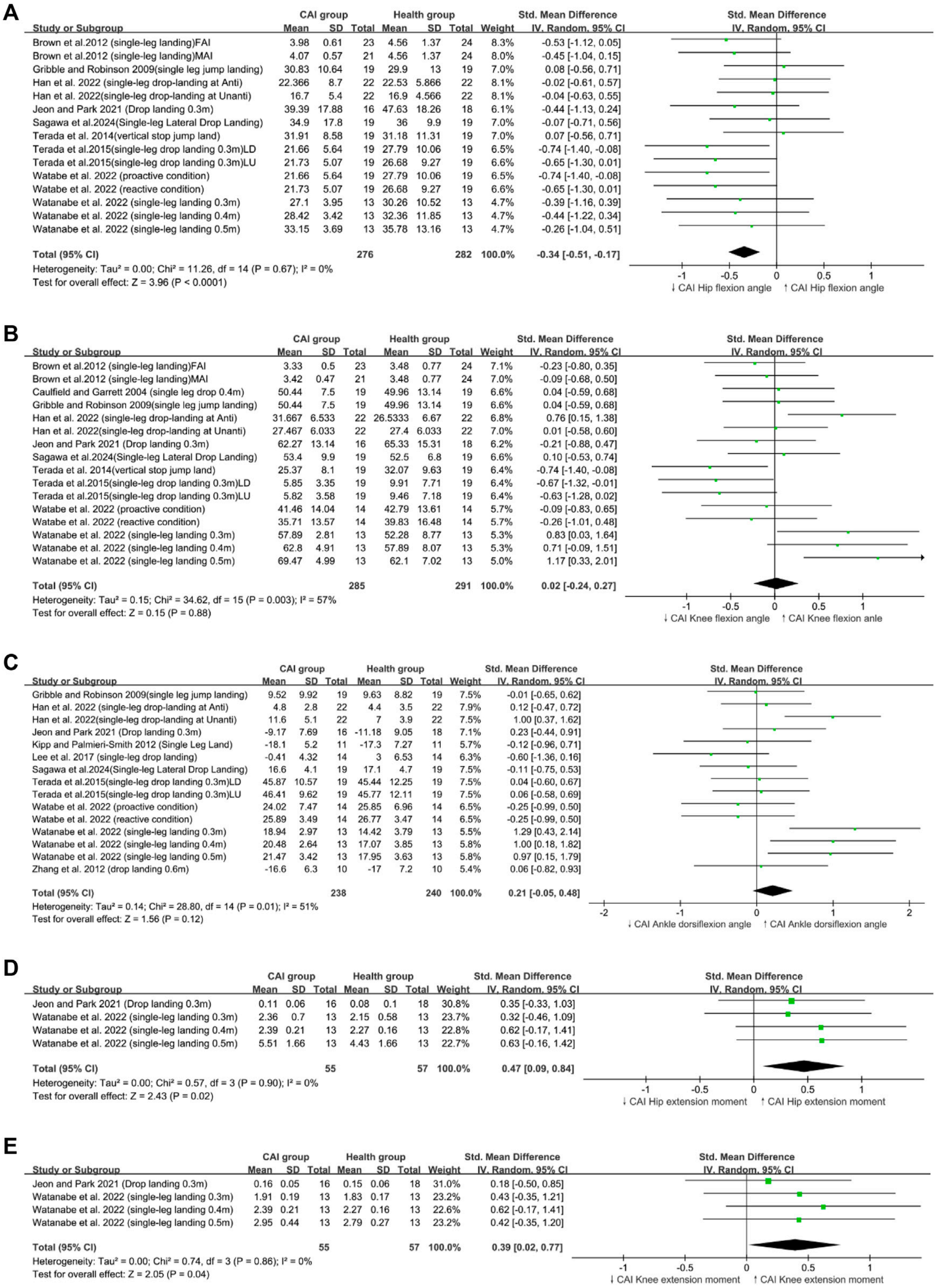

FIGURE 2
Sagittal Plane biomechanical parameters; **(A)** Hip flexion angle, **(B)** Knee flexion angle, **(C)** Ankle dorsiflexion angle **(D)** Hip extension moment, **(E)** Knee extension moment.

knee joint energy dissipation capacity, producing high ACL loads and increasing ACL injury risks (Alentorn-Geli et al., 2009; Norcross et al., 2010).

Overall, the biomechanical pattern of sagittal plane landings in the CAI group, characterized by increased knee and hip extension torques and reduced hip flexion angles, may provide a potential

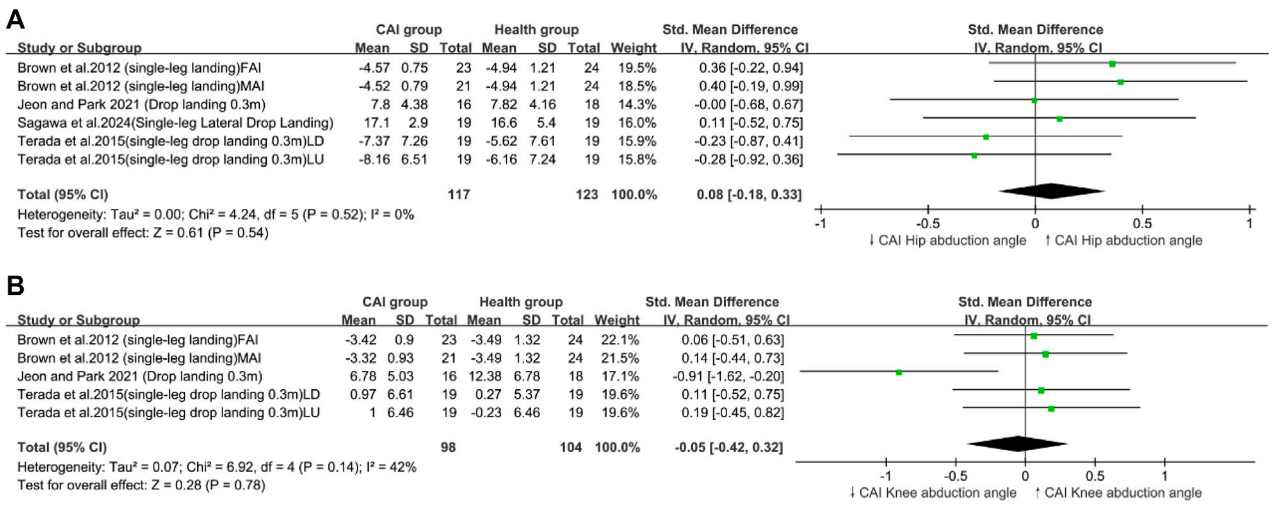

FIGURE 3
Frontal plane kinematics parameters; **(A)**: Hip abduction angle, **(B)**: Knee abduction angle

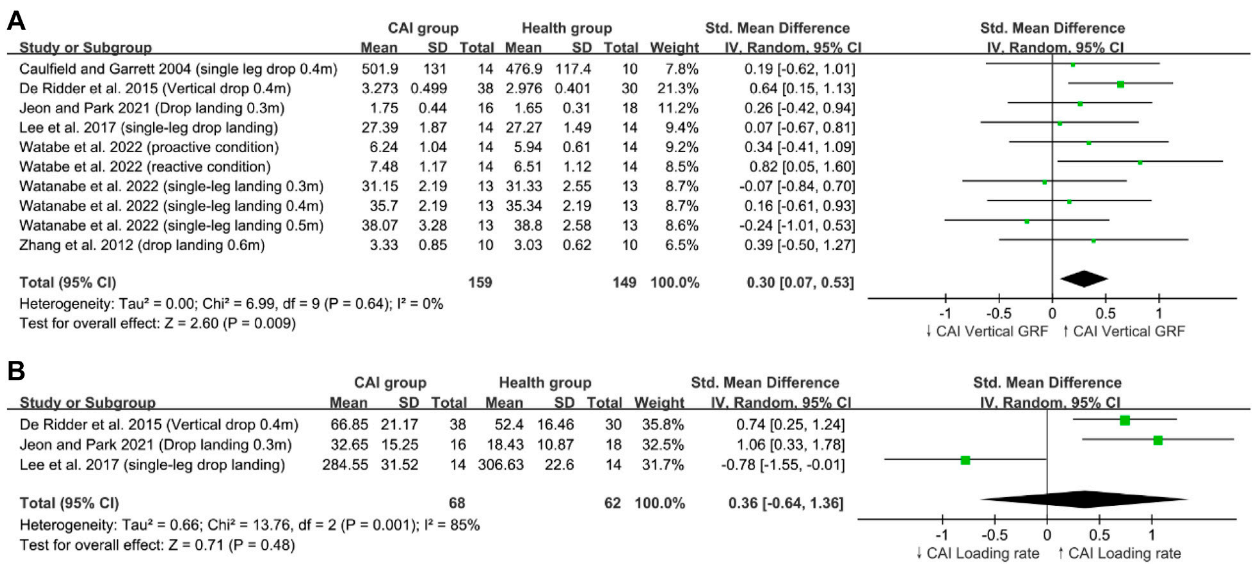

FIGURE 4
Impact loading; **(A)**: Peak VGRF, **(B)**: Loading rate.

mechanism for the heightened risk of individual ACL injuries following CAI.

4.2 Frontal plane kinematics of the lower limb

Contrary to our expectations, no significant differences in hip abduction angles and knee abduction angles were observed between patients with CAI and the healthy group. Our review included four studies on hip abduction angles (Brown et al., 2011; Jeon and Park, 2021; Sagawa et al., 2024; Terada et al., 2015) and three on knee abduction angles (De Ridder et al., 2015; Jeon and Park, 2021; Lee et al., 2017). The collected data generally showed similar knee and hip abduction angles between patients with CAI and healthy individuals during jump-landing tasks, with only one study reporting a reduced knee abduction angle (Jeon and Park, 2021) (Figure 3). Therefore, our meta-analysis found no significant differences between the two groups. During landing tasks, patients with CAI and healthy individuals experienced similar frontal plane kinematics.

Excessive knee abduction angles and knee abduction torques during jump landing are known critical factors influencing ACL injuries (Hewett et al., 2005; Olsen et al., 2004; Quatman and Hewett, 2009). A prospective cohort study of 205 female athletes found that those who suffered ACL injuries had previously exhibited larger knee abduction angles, smaller hip abduction angles, and greater knee abduction torques than their counterparts, These variables were considered predictive of ACL injuries (Hewett et al., 2005). Therefore, avoiding large knee abduction angles and torques and small hip abduction angles during landing may help reduce ACL injury risks in patients with CAI. Given the limited research on related frontal plane dynamics, our study did not include metrics of knee abduction/adduction torques linked to ACL injuries. Future research should incorporate outcomes related to frontal plane biomechanics to elucidate the relationship between the landing frontal plane biomechanics of patents with CAI and ACL injury risks.

4.3 Impact loading

Our review included seven studies on peak VGRF (Caulfield and Garrett, 2004; De Ridder et al., 2015; Jeon and Park, 2021; Lee et al., 2017; Watabe et al., 2022; Watanabe et al., 2021; Zhang et al., 2012) and three on vertical loading rates (De Ridder et al., 2015; Jeon and Park, 2021; Lee et al., 2017) as measures of VGRF and loading rates. The results indicated that patients with CAI exhibited a comparable loading rate with healthy individuals, with an increased peak VGRF (Figure 4).

Previous reviews were consistent with our findings that is, patients with CAI have a high peak VGRF during landing (Jeon et al., 2021; Simpson et al., 2018), which has been identified as a non-contact ACL injury risk (Lin et al., 2012). A prospective study found that female athletes with knee injuries who later experienced ACL ruptures had a 20% higher peak VGRF than those without knee injuries and ACL damage (Hewett et al., 2005; Paterno et al., 2007). The maximum ACL load during landing tasks occurs at the moment of impact peak VGRF (Paterno et al., 2007; Watabe et al., 2022), with an increase in peak VGRF in patients with CAI leading to high ACL loading and great ACL injury risks. Therefore, minimizing the peak VGRF during landing may reduce ACL injury risks in patients with CAI.

The loading rate, as a factor contributing to ACL injuries during landing, has been noted in previous studies to be elevated in patients with CAI (Paterno et al., 2007; Simpson et al., 2018), Although this finding contradicted our results, no significant differences in loading rates were observed. This discrepancy may stem from our dataset, which included only a few studies on loading rates, highlighting the need for further research on this phenomenon.

Other impact loading–related metrics could be used to distinguish between abnormal lower limb impact loads in patients with CAI and healthy individuals. Frontal plane ground reaction forces (Hewett et al., 2009), sagittal plane ground reaction forces (Lin et al., 2012), and symmetry of landing forces are all associated with increased risks of ACL injuries (Paterno et al., 2007; Paterno et al., 2010; Paterno et al., 2011). However, these metrics were not included in this study. In the future, we will to incorporate more indicators of impact loading to clarify the relationship between impact loading during landing in patients with CAI and ACL injury risks.

4.4 Trunk mechanics

Low trunk flexion is associated with ACL injuries (Griffin et al., 2000). A cross-sectional cohort study found that female subjects with ACL injuries exhibited less trunk flexion compared with a control group of women, aligning their trunk with their legs in a way that increases valgus and axial forces (Hewett et al., 2009). High trunk flexion angles can prolong the time to reach peak VGRF, reduce landing impulse, and effectively absorb VGRF, decreasing the load on the ACL (Paterno et al., 2007; Watabe et al., 2022). However, this phenomenon not observed in patients with CAI. This review included two studies on this topic (Brown et al., 2011; Watabe et al., 2022). Watabe et al. (2022) studied the differences in lower limb and trunk biomechanics during active and passive single-leg landings between patients with CAI and healthy individuals, demonstrating that patients with CAI exhibit high trunk flexion angles in active and passive settings. Brown et al. (2011) examined the variability in movements during single-leg landing tasks among FAI, MAI, and healthy groups, and they found no significant differences in trunk flexion angles between patients with FAI/MAI and healthy individuals. Our review included these two studies (Brown et al., 2011; Watabe et al., 2022), and publication bias was statistically tested using Egger's test (p = 0.007), indicating that the likelihood of publishing significant results may be higher than that for nonsignificant results. The presence of this bias necessitates a cautious approach in understanding and interpreting the relationship between trunk flexion and ACL injuries. Future research should utilize a broader range of data sources to ensure the comprehensiveness and balance of the findings. Hence, the clinical significance of the current findings remains uncertain.

Furthermore, our review revealed that patients with CAI exhibited greater trunk lateral flexion than healthy controls (Figure 5). A prospective study of over 900 athletes supported this finding, showing that female athletes with ankle injuries (such as CAI and lateral ankle sprains) demonstrate greater trunk sway compared with uninjured athletes (Beynnon et al., 2006). Insufficient neuromuscular control of the body's core and decreased dynamic postural stability may contribute to increased trunk lateral flexion angles during landing (Kibler et al., 2006). The limitations associated with CAI have been shown to impair dynamic postural stability and neuromuscular control during unilateral jump landings, leading to sensorimotor function impairment (Caulfield et al., 2004; Caulfield and Garrett, 2002; Gribble and Robinson, 2009). Impaired neuromuscular control, especially in the hip muscles, is often observed in patients with CAI, who typically exhibit symptoms of deficient neuromuscular control (Bullock-Saxton et al., 1994; McCann et al., 2017; Terada et al., 2016). Therefore, the increased trunk lateral flexion observed in patients with CAI was due to decreased dynamic postural stability and inadequate neuromuscular control of the hip. Zazulak et al. (2007) suggested that trunk lateral displacement is a strong predictor of knee ligament injury, including ACL injuries. Their study of athletes over 3 years found that those with knee, ligament, or ACL injuries exhibit greater trunk displacement in all planes compared with uninjured athletes, identifying lateral displacement as the strongest predictor of ligament injuries. Trunk lateral flexion causes the lateral ground reaction force vector to shift sideways and, having a longer lever arm relative to the knee joint center, directly increases the likelihood of knee abduction motion and torque, thereby elevating the risk of ACL injuries (Hewett et al., 2009). Therefore, the landing mechanics of patients with CAI characterized

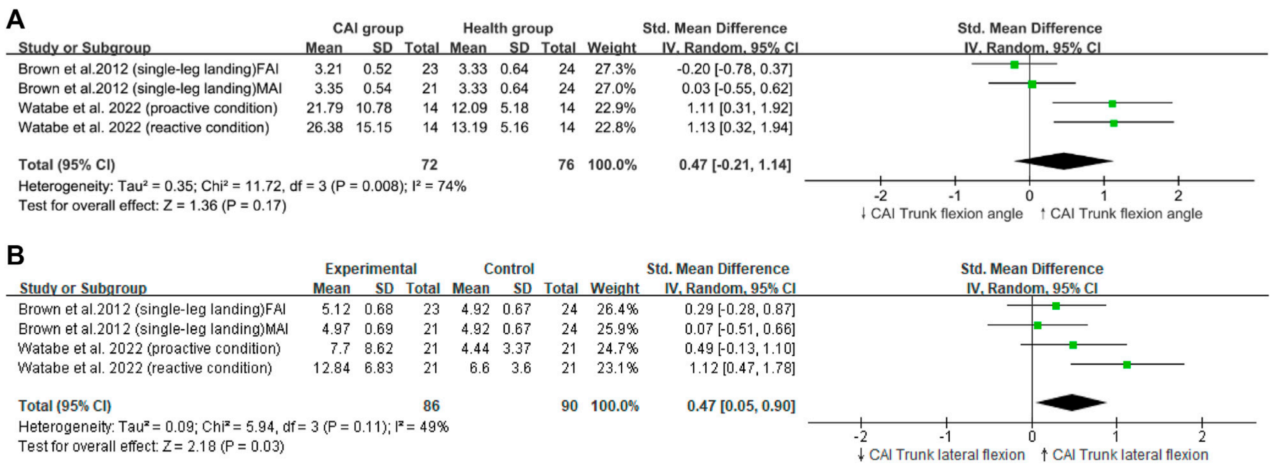

FIGURE 5
Trunk mechanics; **(A)**: Trunk flexion, **(B)**: Trunk lateral flexion.

by pronounced trunk lateral flexion could be factor contributing to the increased risk of ACL injuries in this population.

4.5 Limitation

Our study has the following limitations and weaknesses. First, the current research includes young athletes (under 27 years old) from various sports and regularly exercising individuals. Therefore, our results may not be applicable to other populations with different ages and levels of physical activity. Second, given limited previous research on horizontal biomechanical variables related to ACL injuries and knee coronal plane torque as outcome measures, we were unable to perform meta-analyses on these aspects. Third, many risky movements,such as cutting and sudden stopping, can lead to ACL injuries; considering the heterogeneity and representativeness of injury movements, we selected only landing actions for analysis. Fourth, our results only indicate the differences in landing biomechanics between individuals with CAI and healthy controls during landing tasks. We cannot ascertain whether these differences would vary with different experimental protocols. Even for the same type of jump-landing tasks, the implementation protocols were not uniform. We also observed variations in platform height and single versus double foot landings, which led to high heterogeneity in some outcome indicator analyses. Lastly, each meta-analysis included only a small number of studies; Thus, the results of this systematic review should be interpreted with caution.

5 Conclusion

This study confirmed the association between CAI and increased sagittal plane kinetics, reduced hip flexion angles, increased peak VGRF, and increased trunk lateral flexion, all of which are related to a heightened risk of ACL injuries during landing tasks. These findings provide a basis for improving the understanding of ACL injury risks during landing tasks in individuals with CAI. This knowledge can guide future preventative measures and rehabilitation strategies to mitigate ACL injury risks 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.

Author contributions

ZH: Formal Analysis, Investigation, Methodology, Visualization, Writing–original draft, Writing–review and editing. HZ: Funding acquisition, Project administration, Supervision, Validation, Visualization, Writing–review and editing. BY: Investigation, Methodology, Writing–original draft, Writing–review and editing. ZZ: Data curation, Validation, Writing–review and editing. GL: Data curation, Validation, Writing–review and editing. HP: Supervision, Writing–review and editing. RL: Supervision, Validation, Visualization, Writing–review and editing.

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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.2024.1428879/full#supplementary-material

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Influences of fatigue and anticipation on female soccer players' biomechanical characteristics during 180° pivot turn: implication for risk and prevention of anterior cruciate ligament injury

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Introduction: Athletes' capability to perform activities with body rotation could be weakened by fatigue accumulation. Making pivot turning in unanticipated scenarios after fatigue may greatly challenge athletes' ability to adapt rational motion strategies, elevating the risk of anterior cruciate ligament (ACL) injury. This study aimed to investigate the effects of fatigue and anticipation on biomechanical risk factors of ACL injury during 180° pivot turns in female soccer players.

Methods: Twenty-one female soccer players were selected as participants. The participants performed anticipated turning maneuver before the fatigue intervention. The participants sprinted along the runway, decelerated and planted their foot on the force plate, and then executed a 180° pivot turn. For unanticipated tests, the pivot turn was mixed with side/cross-cuts, which were indicated to the participant using a custom-designed light system. The tests were repeated by the participant after receiving a fatigue intervention. Lower-limb joint angles and moments were characterized. Peak ground reaction forces (GRFs) and GRF loading rates were determined. Two-way repeated measures analysis of variance was applied to examine the effects of fatigue and anticipation on the variables of interest.

Results: Compared to the anticipated conditions, the approach speed was significantly lower in the unanticipated tests ($P < 0.0001$). Lower-limb kinematics showed varied angular patterns across conditions: greater hip joint variations in flexion, abduction, and internal rotation during unanticipated turns; consistent knee joint flexion and ankle plantarflexion with dorsiflexion observed mid-turn. Significant interactions ($P = 0.023$ to $P = 0.035$) between fatigue and anticipation influenced hip joint angles. Anticipation effects were notable at initial contact and peak ground reaction force, increasing hip, knee, and ankle joint angles ($P < 0.0001$ to $P = 0.012$). Participants showed consistent ground reaction force (GRF) patterns during pivot turns across fatigue and anticipation conditions, with the first peak occurring approximately 10% into the turn period. Significant interaction effects (P = 0.016) between fatigue and anticipation were observed for knee flex/extension moments at the first peak vertical GRF. Anticipation significantly increased first peak vertical (P < 0.0001), anteroposterior (P < 0.0001), and mediolateral (P < 0.0001) GRFs. Fatigue increased first peak vertical (P = 0.022), anteroposterior (P = 0.018), and mediolateral (P = 0.019) GRFs. Post-fatigue, participants exhibited reduced first peak GRFs and loading rates compared to pre-fatigue conditions, with higher rates observed in unanticipated turns (vertical GRF: P = 0.030; anteroposterior GRF: P < 0.0001).

Conclusion: Female soccer players' lower-limb Biomechanical characterization could be greatly affected by the change of anticipatory scenarios. With the associated increase of GRF, the risk of their ACL injury might be elevated. Fatigue affected female soccer players' abilities on movement performances, but the interaction of these two factors could potentially weaken their knee's functions during pivot turns. Cognitive training on unanticipated tasks may be important for rehabilitation training after ACL injury.

KEYWORDS

anterior cruciate ligament injury, fatigue, anticipation, turn, lower-limb biomechanics

Introduction

Epidemiological studies have shown that non-contact anterior cruciate ligament (ACL) injury is prevalent in the context of soccer matches (Boden et al., 2000; Collins et al., 2016). It was reported that approx. 70% of the ACL injuries in female soccer players occurred in the braking phase of pivot turns during competitions (Faude et al., 2005). Each player could complete over 700 times of movements with directional change in a soccer match (Nygaard Falch et al., 2019), which is often rapid in the circumstance of evading opponents or defense. From the biomechanical perspective, this may be attributed to the fact that knee joints possess great transversal rotations during pivot turn, with concomitant high loadings acted on the knee during body deceleration (Collins et al., 2016; Dos' Santos et al., 2019; McLean et al., 2004). Therefore, investigating individuals' lower-limb biomechanics in turn maneuvers is crucial for understanding the mechanisms of non-contact ACL injuries in soccer, particularly for the development of prevention and rehabilitation training programs.

Soccer players accumulate fatigue and encounter unanticipated scenarios during competition. Fatigue and anticipation have also been reported to greatly affect an individual's lower-limb kinematics and kinetics during movements with directional change such as single-step cuts (Collins et al., 2016; Borotikar et al., 2008; McLean and Samorezov, 2009). However, the findings from previous studies are controversial. Some studies reported that fatigue and anticipation altered knee mechanics towards an increased risk of ACL injury in cut maneuvers (Collins et al., 2016; Borotikar et al., 2008). Other studies claimed that individuals adopted a protective strategy during post-fatigue (Cortes et al., 2014) or unanticipated cuts (Rolley et al., 2023), implying a potential reduction of ACL loading. Importantly, cut and pivot turn are very much different tasks, with dissimilar motion characteristics and demands (Greig, 2009). While individuals decelerate their body and change their direction simultaneously in a cut maneuver, the pivot turn could consist of a more prolonged and intensive braking phase before any body rotation happens. Therefore, people have argued that cut maneuvers cannot replicate the demands of a pivot turn which could more realistically represent a soccer task (Greig, 2009). Furthermore, a previous study has reported that compared to lateral cutting and landing movements, individuals exhibited smaller knee flexion and larger abduction at peak ground reaction forces (GRF) during pivot turns, with a higher knee joint loading that might potentially increase the risk of ACL injury (Cortes et al., 2011). Thus, individuals' performances in pivot turns under neuromuscular perturbations cannot be fully explained by the findings from cutting tests, and should be investigated separately.

There is currently a lack of clear description on the influences of fatigue and anticipation to soccer players' lower-limb biomechanics in pivot turn. During pivot turning, individuals may utilize complicated lower-limb movement coordination strategies to counteract the forward momentum of their upper body, and to connect with the subsequent body rotation (Newell, 1996; Newell and Slifkin, 1998). This substantially challenges the ability of their lower-limb joints, especially the knee, to resist overlarge rotations and loadings. Therefore, any fatigue on the lower-limb muscles may not only affect the muscular protection to the knee's stability, but also change the overall motion control in the turn maneuvers. A previous study reported that soccer players could exhibited either protective or dangerous biomechanical behaviors during pivot turns after fatigue (Zago et al., 2021). Such inconsistency in their biomechanical performance may be explained by the compromise of the central and peripheral processing mechanisms under fatigue (Borotikar et al., 2008; Lorist et al., 2005). Importantly, when players are facing unanticipated tasks in a fatigue situation, they may likely experience the worst-case scenario for coordination of central and peripheral responses compared to either of these two situations alone. Considering fatigue and decision making in unanticipated task are commonly integrated in soccer matches, it is essential to investigate their interactions on the soccer players' biomechanical performances in pivot turns, especially those factors associated ACL injury. Nevertheless, few studies have achieved this goal till now.

Studies have shown that female soccer players face 3–5 times higher risk of ACL injuries than their male counterparts, and this difference is largely attributed to differences in physiology and movement technique (Prodromos et al., 2007; Moses et al., 2012). Therefore, the purpose of this study was to test female soccer players' biomechanical performances during anticipated and unanticipated turns, with a special focus on those variables matching the known ACL injury mechanism. This study also aimed to repeat the measurements after the participants received a fatigue drilling intervention, in order to study the integrated effects of anticipation and fatigue on their performances. The hypotheses of this study were: 1) Anticipation would greatly influent the kinematics and kinetics of the participants' lower-limb joints in turn maneuvers. 2) The simultaneous presence of fatigue would further amplify their biomechanical changes, potentially resulting in even a higher risk of ACL injury.

Methods

Sample size calculation

The sample size calculation was performed using G*Power 3.0.10 software (Heinrich-Heine-Universität Düsseldorf, Germany). An effect size f of 0.27, derived from a pilot study, was used. The calculation aimed for a power of 0.80 and an alpha level of 0.05. The statistical method used for the calculation was a repeated measures ANOVA with within-subject factors. It was determined that a minimum of 21 participants would be required to detect significant effects.

Participants

Participants were 21 female college soccer players, all of whom had a minimum of 5 years of training experience at the college level. All participants were national level athletes or above and had finished in the top three in national competitions. This high level of training background means that there may be significant differences in their motor skills, body control, and motor coping abilities, which is crucial for the interpretation of the results of the study of the effects of fatigue and anticipation on their motor performance. Recruitment was conducted through the professional networks of the research team members. The inclusion criteria included: 1) right dominant legs; 2) no cardiovascular or respiratory disease; 3) no history of lower limb surgery; 4) no injury to lower limbs in the past 6 months. This study was approved by the local Ethics Committee. Before measurement, participants were informed of the study aims and experiment protocol and then signed the informed consent form.

Experimental setup

A 16-camera motion analysis system (Vicon Motion Analysis Inc., Oxford, United Kingdom; 200 Hz) and a force plate (Kistler Instruments AG, Winterthur, Switzerland; 1,000 Hz) were used to capture 3D body motions and GRFs. Fourteen-millimeter reflective markers were bilaterally placed on specific anatomical landmarks, including the head of the first and fifth metatarsi, calcaneus, medial and lateral malleolus, medial and lateral femoral epicondyles, greater trochanter of the femur, anterior superior iliac spine, iliac crest, sacrum, sternum, xiphisternal joint, and acromion. Sixteen tracking markers were placed on the participant's bilateral shanks and thighs. The test site is shown in Figure 1.

Overall experiment protocol

Participants wore a spandex T-shirt, shorts, and sports shoes during the experiment. After 5-min warming up on a treadmill at 8 km/h, the participant was prepared for optical motion capture (MoCop) tests. The participant performed unanticipated and anticipated pivot turn tests (see section "Pivot turn test"), while their 3D body motions and GRFs were captured. The tests were performed in a randomized order with 30 s rest between the trials. Afterwards, the participant received fatigue interventions (see section "Fatigue protocol"). Once the target level of fatigue was met, the participant immediately repeated the unanticipated and anticipated turn tests.

Fatigue protocol

The fatigue protocol consisted of a variety of movements including running, sidestep cutting, and jumping. Participants performed three countermovement jumps (CMJs) at a rhythm of 18 beats per minute regulated by a metronome. The peak jumping height calculated from the vertical displacements of a reflective marker on the participant's sacrum was used as a reference to quantify their level of fatigue (Zago et al., 2021). Participants then stood in the center of a circular region with a diameter of 6 m, where six light buttons were placed on the circle's edge. The buttons illuminated in a randomized order, and participants moved at their best to press the illuminated button, completing 25 presses followed by three CMJs. This cycle was repeated with 10-s breaks until a 30% reduction in jumping height was achieved, indicating the target level of fatigue. Participants spent an average of 39.2 ± 5.3 min performing 12.9 ± 3.5 fatigue drill cycles to achieve the target level of fatigue.

Pivot turn test

Participants performed both anticipated and unanticipated 180° pivot turn tests on a 9-m-long and 2-m-wide runway with a force plate embedded at its center. An infrared barrier system was placed 2 m before the force plate to accurately capture the moment of initial contact (IC). In the anticipated condition, participants were informed of the movement direction (pivot turn) beforehand and instructed to execute the turn using their right leg.

In the unanticipated condition, cutting movements were introduced to mimic the unpredictable nature of real-game scenarios. Upon reaching the infrared barrier system, a custom-designed light system was triggered, indicating the required movement direction just before the participants reached the force plate. The direction was signaled using arrows: a left arrow for a side

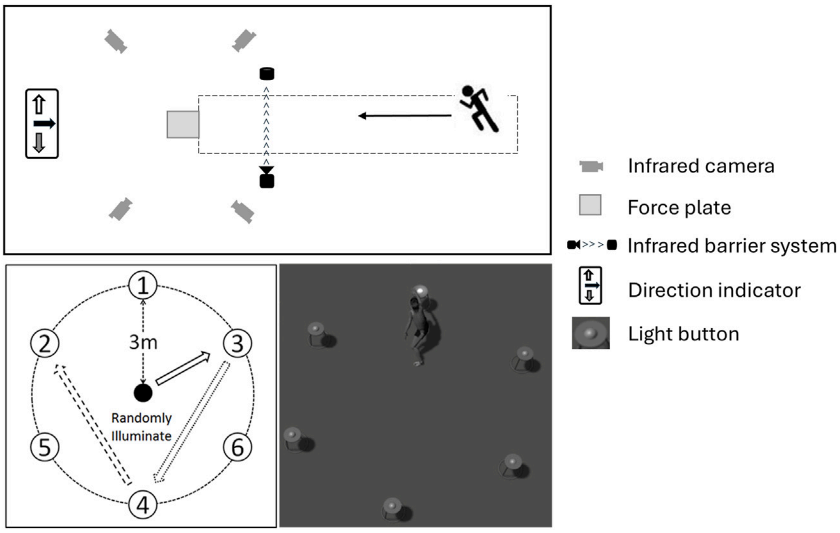

FIGURE 1
Illustration of pivot turn test and fatigue intervention protocol. Upper: The setup of the pivot test; bottom left: Scheme of participant's movement directions during the fatigue intervention; bottom right: Illstration of the fatigue intervention.

cut (45°), a right arrow for a cross-cut (90°), and a downward arrow for a pivot turn (180°). The order of light gate activation was randomized using a random number generator (www.random.org), ensuring that participants could not prepare in advance.

To perform the pivot turn, participants had to decelerate from their sprint, plant their right foot on the force plate, and execute a 180° turn using their right leg as the pivot. A test was considered successful if the participant completed the turn within the designated area without stepping outside the force plate, maintained balance throughout the motion, and executed the turn fluidly without hesitation or excessive pausing. Trials were repeated if these criteria were not met to ensure consistency and reliability in the collected performance data. Three successful trials were recorded for the unanticipated and anticipated conditions.

Data processing

The biomechanical analysis suite, Visual3D (C-Motion Inc., Germantown, MD, United States), was used to compute participants' 3D kinematic and kinetic variables. Marker trajectories and analog signals from the force plate were low-pass filtered using a Butterworth 4th-order bidirectional filter with cutoff frequencies of 20 Hz and 100 Hz, respectively. The 3D angular variables were defined using a Cardan sequence (X-Y-Z), with the rotation order as flexion/extension (*X*-axis), abduction/adduction (*Y*-axis), and internal/external rotation (*Z*-axis). The GRF was expressed in three directions: vertical (V-GRF), anteroposterior (AP-GRF), and mediolateral (ML-GRF). A right-hand rule was applied to determine the polarity of the joint angles and GRFs.

The pivot turn period was defined from the IC of the participant's foot on the force plate (vertical force over 10N) to when the foot left the force plate (vertical force below 10N). Approach speed was calculated as the sprinting speed at IC. Lower-limb joint angles and GRFs throughout the pivot turn period were time-normalized as a percentage of the turn period and then plotted. The first peak GRFs in the three directions were determined and normalized to each participant's body weight (BW). The GRF loading rate was calculated as the quotient of the first peak V-GRF and the time to reach this peak from IC.

Knee joint moments at the first peak V-GRF were characterized and normalized to each participant's body mass (kg). Joint angles of the hip, knee, and ankle at IC were extracted, and knee joint angles at the first peak V-GRF were characterized. Joint angles and moments in the sagittal, frontal, and transverse planes were reported to provide a comprehensive analysis of joint movements and their potential relevance to non-contact ACL injuries.

Joint angles at IC and peak GRF were selected based on their relevance to ACL injury mechanisms, as indicated by previous studies (Zago et al., 2021). Initial contact is a critical phase where the body's alignment and initial load distribution can influence injury risk. The first peak V-GRF represents the maximum vertical force experienced, capturing the peak loading conditions that are critical for understanding knee mechanics and potential injury mechanisms during pivot turns. Analyzing these points provides insights into the biomechanical demands and injury risk factors associated with these specific moments during dynamic movements.

Statistical analysis

The results were averaged from outcomes of the three trials in each condition and were presented as mean ± standard deviation (SD). Repeated measures analysis of variance (ANOVA) was conducted to evaluate the kinematic and kinetic variables of interest, with the factors being fatigue (pre-fatigue vs. post-fatigue) and anticipation (anticipated vs. unanticipated). Assumptions of normality and sphericity for Repeated Measures ANOVA were examined. Normality was assessed using the Shapiro-Wilk test, and sphericity was evaluated using Mauchly's test. If the

TABLE 1 Approach speed before the pivot turn motion.

Variables	Pre-fatigue		Post-fatigue		Fatigue	Anticipation	Interaction
	Anticipated	Unanticipated	Anticipated	Unanticipated	*P*	*P*	*P*
Approach speed (m/s)	2.60 ± 0.26	2.38 ± 0.22	2.53 ± 0.19	2.37 ± 0.26	0.100	**<0.001***	0.336

*Bold value indicates statistical significance ($P < 0.05$).

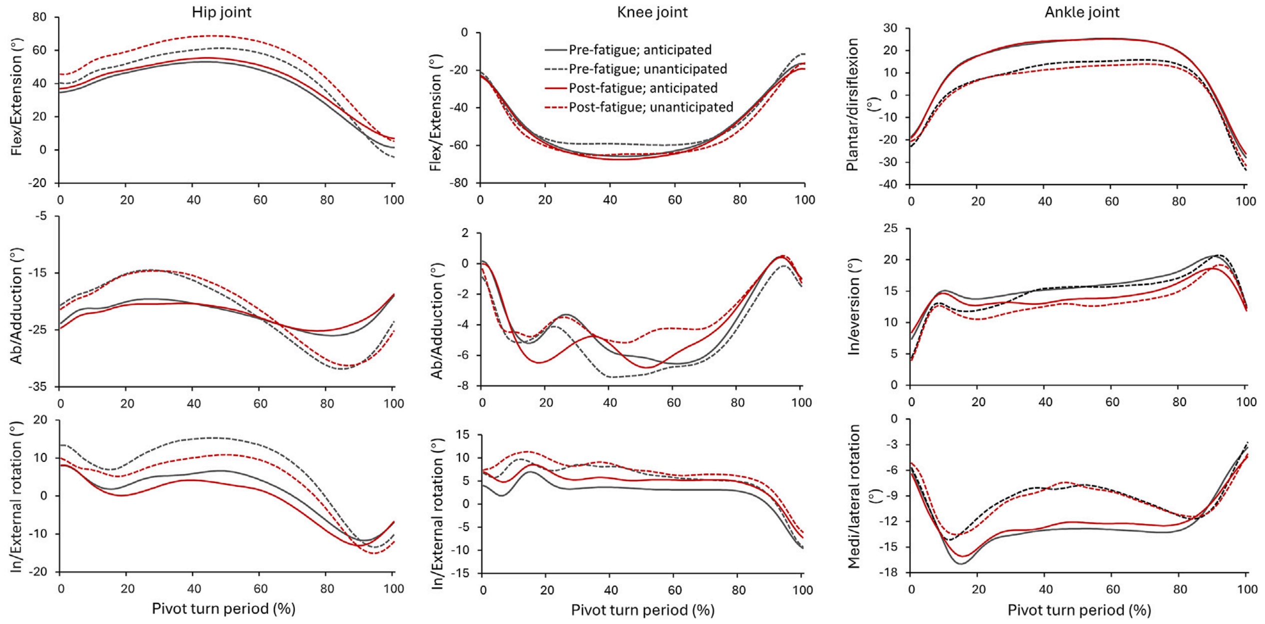

FIGURE 2
Angular variations of hip, knee, and ankle joints throughout the pivot turn period in unanticipated and anticipated tests before and after fatigue. The joint angles were expressed in three planes: sagittal (flex/extension), frontal (ab/adduction), and transverse (in/external rotation) planes.

assumption of sphericity was violated, the Greenhouse-Geisser correction was applied. In cases where the assumption of normality was violated, the non-parametric Aligned Rank Transform test was used as an alternative. When significant interactions between fatigue and anticipation were found, simple main effects were analyzed to determine the specific effects of anticipation at each level of fatigue and the effects of fatigue at each level of anticipation. Effect size (ES) was calculated using the method of Cohen's d and classified as: small 0.2–0.49, medium 0.5–0.79, and large >0.8 (Yeung, 2014). Statistical analysis was performed using SPSS software (version 28.0; SPSS Inc., Chicago, IL, United States). An alpha level of 0.05 was set for all statistical tests.

Result

Participant demographics

The demographic data of the participants are as follows: age: 18.4 ± 1.4 years; height: 1.66 ± 0.05 m; mass: 60.1 ± 5.5 kg; training experience: 9.2 ± 1.6 years. Initially, 21 participants were recruited, all of whom completed the study with no dropouts.

Approach speed

Table 1 presents the approach speeds before the pivot turn motion. There was no significant interaction between fatigue and anticipation on approach speed (Table 1). However, a significant main effect of anticipation was found ($F_{1,20} = 54.241$, $P < 0.001$, d = 0.812), with participants sprinting significantly slower in unanticipated turns compared to anticipated conditions. No significant main effect of fatigue was observed.

Lower limb kinematics

The angular patterns of the lower-limb joints in the pivot turns under different conditions were shown in Figure 2. The participant's hip joint possessed flexion and abduction during the entire turn period, with a change from internal to external rotation. It could be observed that compared to their performance during anticipated turns, the participants exhibited greater angular variations in their hip joints in all the three anatomical planes during unanticipated turns. In addition, the participant's knee joint possessed flexion, abduction, and internal rotation (0% to approx. 95%) during the turn period. Their knee angular variations were similar among

TABLE 2 Lower-limb joint kinematics at IC and the first peak ground reaction force in pre- and post-fatigue states during anticipated and unanticipated pivot turns.

Time point	Variables	Pre-fatigue		Post-fatigue		Fatigue	Anticipation	Interaction
		Anticipated	Unanticipated	Anticipated	Unanticipated	*P*	*P*	*P*
IC	Hip flex/extension angle (°)	34.7 ± 11.1	40.3 ± 18.8	36.9 ± 11.3	45.6 ± 12.9	0.164	**0.002***	0.469
	Hip ab/adduction angle (°)	−23.8 ± 7.6	−20.6 ± 6.6	−24.7 ± 7.1	−21.3 ± 6.1	0.365	**0.002***	0.914
	Hip in/external rotation angle (°)	8.0 ± 7.9	13.4 ± 8.6	8.1 ± 5.5	9.9 ± 7.1	0.123	**0.001***	**0.023***
	Knee flex/extension angle (°)	−22.6 ± 6.2	−20.9 ± 10.9	−23.2 ± 5.1	−22.8 ± 5.0	0.265	0.378	0.594
	Knee ab/adduction angle (°)	0.2 ± 4.7	−0.9 ± 4.6	−0.0 ± 4.1	−0.4 ± 4.3	0.777	0.140	0.311
	Knee in/external rotation angle (°)	4.0 ± 8.8	6.7 ± 8.6	7.0 ± 9.2	7.4 ± 8.5	0.222	0.166	0.178
	Ankle plantar/dorsiflexion angle (°)	−18.6 ± 8.9	−22.8 ± 8.3	−19.1 ± 6.6	−20.6 ± 7.2	0.535	**0.004***	0.052
	Ankle in/eversion angle (°)	7.4 ± 6.6	4.4 ± 6.3	8.4 ± 6.3	4.0 ± 7.3	0.712	**<0.001***	0.062
	Ankle medi/lateral rotation angle (°)	−5.9 ± 4.4	−5.7 ± 4.8	−6.4 ± 5.4	−5.2 ± 4.8	0.967	0.158	0.186
First peak vGRF	Hip flex/extension angle (°)	37.4 ± 10.7	41.4 ± 18.4	40.0 ± 11.9	48.2 ± 13.6	0.087	**0.009***	0.347
	Hip ab/adduction angle (°)	−21.5 ± 8.5	−18.8 ± 7.3	−22.3 ± 7.5	−19.4 ± 6.9	0.435	**0.004***	0.898
	Hip in/external rotation angle (°)	3.9 ± 7.7	10.9 ± 7.8	3.4 ± 6.8	7.3 ± 7.2	0.101	**<0.001***	**0.035***
	Knee flex/extension angle (°)	−35.3 ± 8.1	−30.7 ± 12.8	−36.6 ± 7.3	−33.9 ± 5.2	0.073	**0.012***	0.476
	Knee ab/adduction angle (°)	−3.4 ± 4.1	−3.6 ± 4.7	−3.7 ± 4.1	−4.1 ± 4.6	0.461	0.544	0.898
	Knee in/external rotation angle (°)	0.2 ± 6.9	4.4 ± 7.5	2.5 ± 5.5	7.6 ± 6.0	0.058	**<0.001***	0.456
	Ankle plantar/dorsiflexion angle (°)	1.9 ± 12.0	−9.9 ± 7.8	2.4 ± 10.9	−10.0 ± 7.9	0.889	**<0.001***	0.665
	Ankle in/eversion angle (°)	13.8 ± 6.5	10.7 ± 6.6	14.0 ± 5.8	11.4 ± 6.1	0.576	**<0.001***	0.695
	Ankle medi/lateral rotation angle (°)	−10.7 ± 6.3	−9.4 ± 6.0	−10.4 ± 5.8	−9.0 ± 5.5	0.731	**0.005***	0.911

IC, initial contact; vGRF, vertical ground reaction force, *Bold values indicate statistical significance ($P < 0.05$).

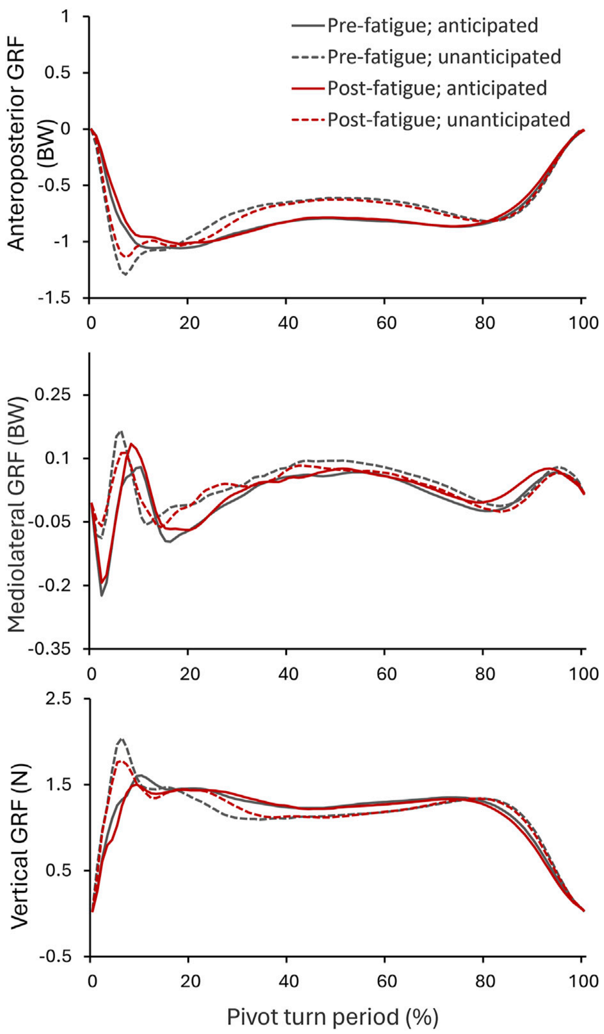

FIGURE 3
Variations of ground reaction forces throughout the pivot turn period in unanticipated and anticipated tests before and after fatigue. The ground reaction forces were expressed in three directions: anteroposterior, mediolateral, and vertical vectors.

different fatigue and anticipation conditions. Finally, they exhibited ankle plantarflexion during the first and last 10% of the turn period, but dorsiflexion during the middle phase. Meanwhile, they exhibited ankle inversion and lateral rotation. They ankle angular variations in the sagittal and transverse planes were smaller during unanticipated turns compared to anticipated turns.

Table 2 details the kinematics of lower-limb joints at IC and at the first peak vertical ground reaction force (vGRF). Significant interactions between fatigue and anticipation were observed for hip in/external rotation angle at IC ($F_{1,20}$ = 6.075, P = 0.023, d = 0.160) and at first peak vGRF ($F_{1,20}$ = 5.017, P = 0.035, d = 0.163). For the main effects, anticipation significantly affected multiple variables at IC. During unanticipated turns, the participants exhibited significantly greater hip flexion angle ($F_{1,20}$ = 12.301, P = 0.002, d = 0.518), smaller hip abduction angle ($F_{1,20}$ = 13.025, P = 0.002, d = 0.477), greater hip internal rotation angle ($F_{1,20}$ = 16.634, P = 0.001, d = 0.488), greater ankle plantarflexion angle ($F_{1,20}$ = 10.393, P = 0.004, d = 0.359), and smaller ankle inversion angle ($F_{1,20}$ = 32.434, P < 0.001, d = 0.556). At the first peak vGRF, the participants possessed significantly larger hip flexion angle ($F_{1,20}$ = 8.299, P = 0.009, d = 0.437), smaller hip abduction angle ($F_{1,20}$ = 10.425, P = 0.004, d = 0.370), larger hip internal rotation angle ($F_{1,20}$ = 22.939, P < 0.001, d = 0.488), smaller knee flexsion angle ($F_{1,20}$ = 7.606, P = 0.012, d = 0.420), larger knee internal rotation angle ($F_{1,20}$ = 34.694, P < 0.001, d = 0.707), larger ankle plantarflexion angle ($F_{1,20}$ = 60.131, P < 0.001, d = 1.237), smaller ankle inversion angle ($F_{1,20}$ = 30.575, P< 0.001, d = 0.444), and smaller ankle lateral rotation angle ($F_{1,20}$ = 10.024, P = 0.005, d = 0.235).

Lower limb kinetics

The patterns of GRF during the pivot turns under different conditions were shown in Figure 3. The GRF variations were similar among different fatigue and anticipation conditions. The first peak GRF occurred shortly after IC, at approx. 10% of the turn period.

Table 3 provides detailed information on GRFs and knee joint moments. A significant interaction between fatigue and anticipation was found for the knee flex/extension moment at first peak vertical GRF ($F_{1,20}$ = 6.979, P = 0.016, d = 0.686). Simple effects analysis indicated that before the fatigue intervention, the knee flexion moment was significantly greater during unanticipated turns compared to anticipated turns (P = 0.006), but this difference was not significant after the fatigue intervention (P = 0.154). In terms of changes in joint kinetics, the main effect of anticipation was signficiant on knee ab/adduction and in/external rotation moment at first peak vertical GRF. Compared to their perforamces during anticipated turns, the participant exhibitied significantly smaller knee adduction moment ($F_{1,20}$ = 54.831, P < 0.001, d = 1.244) and smaller internal rotation moment ($F_{1,20}$ = 39.211, P < 0.001, d = 1.358) during unanticipated turns.

For main effects, anticipation significantly affected the first peak vertical GRF ($F_{1,20}$ = 27.996, P < 0.001, d = 0.989), first peak anteroposterior GRF ($F_{1,20}$ = 48.128, P < 0.001, d = 0.883), and first peak mediolateral GRF ($F_{1,20}$ = 32,621, P < 0.001, d = 0.232). Fatigue showed significant effects on the first peak vertical GRF ($F_{1,20}$ = 6.147, P = 0.022, d = 0.337), first peak anteroposterior GRF ($F_{1,20}$ = 6.644, p = 0.018, d = 0.318), and first peak mediolateral GRF ($F_{1,20}$ = 6.521, P = 0.019, d = 0.103). The main effects of fatigue and anticipation on the loading rate of GRF were also significant. The loading rate to the first peak vertical GRF was signficantly smaller in post-fatigue turns than that in pre-fatigue turns ($F_{1,20}$ = 5.478, P = 0.030, d = 0.336), and was significantly larger in unanticipated turns than that in anticipated turns ($F_{1,20}$ = 6.245, P = 0.002, d = 0.584). Similarly, the loading rate to the first peak posterior GRF was significantly larger during unanticipated turns compared to anticipated turns ($F_{1,20}$ = 18.569, P< 0.001, d = 0.835).

Discussion

Concurrence fatigue and unanticipated tasks can cause deviations in soccer players' performances during real compitations (Rolley et al., 2023; Benjaminse et al., 2019). Their integrative influence on individuals' lower-limb kinematics and kinetics in rapid pivot turns, especially those associated with

TABLE 3 Ground reaction forces and knee joint moments in pre- and post-fatigue states during anticipated and unanticipated pivot turns.

Variables	Pre-fatigue		Post-fatigue		Fatigue	Anticipation	Interaction
	Anticipated	Unanticipated	Anticipated	Unanticipated	*P*	*P*	*P*
First peak vertical GRF (BW)	2.26 ± 0.39	2.76 ± 0.44	2.17 ± 0.48	2.55 ± 0.45	**0.022***	**<0.001***	0.181
First peak anteroposterior GRF (BW)	−1.33 ± 0.22	−1.53 ± 0.21	−1.27 ± 0.20	−1.45 ± 0.23	**0.018***	**<0.001***	0.710
First peak mediolateral GRF (BW)	−0.27 ± 0.11	−0.19 ± 0.10	−0.24 ± 0.09	−0.14 ± 0.76	**0.019***	**<0.001***	0.623
Loading rate of vertical GRF (BW/s)	73.37 ± 35.76	109.15 ± 51.08	68.48 ± 38.61	87.33 ± 30.34	**0.030***	**0.002***	0.150
Loading rate of anteroposterior GRF (BW/s)	27.52 ± 15.94	38.48 ± 14.55	23.15 ± 17.33	38.66 ± 15.39	0.282	**<0.001***	0.193
Knee flex/extension moment (N•m/kg)	−0.00 ± 1.20	−0.83 ± 1.32	−0.59 ± 1.13	−0.46 ± 1.24	0.432	0.193	**0.016***
Knee ab/adduction moment (N•m/kg)	2.82 ± 0.82	1.56 ± 1.16	2.56 ± 0.89	1.46 ± 0.87	0.235	**<0.001***	0.562
Knee in/external rotation moment (N•m/kg)	0.49 ± 0.20	0.22 ± 0.26	0.50 ± 0.19	0.21 ± 0.22	0.978	**<0.001***	0.710

BW, body weight; GRF, Ground Reaction Force; *Bold values indicate statistical significance ($P < 0.05$).

non-contact ACL injury, however remain less known. This study characterized soccer players' biomechanical performances during anticipated and unanticipated pivot turns, both before and after fatigue. In general, this study revealed that anticipation affected the participants' lower-limb biomechanics towards an increased risk of ACL injury during pivot turning. The results also indicated that the participants' movement abilities were weakened by fatigue. Finally, this study revealed significant interactions between fatigue and anticipation factors particularly for the hip and knee biomechanical performances, underscoring the complex interplay of biomechanical adjustments athletes undergo to cope with cognitive challenges during turning tasks.

Kinematic analysis

Participants exhibited pronounced changes in hip joint kinematics, particularly in unanticipated turns. Specifically, there was increased hip flexion and abduction angles, alongside greater variability in internal and external rotation angles. This suggests that unanticipated situations necessitate rapid and extensive hip movements to accommodate abrupt changes in movement demands. Moreover, increased hip internal rotation at IC during unanticipated turns is consistent with previous research (Clemens and Pew, 2023; Guex et al., 2012). Such internal rotation is associated with heightened knee joint abduction moments during directional changes (McLean et al., 2005), potentially compromising the efficacy of quadriceps and hamstrings in countering external knee abduction forces (Delp et al., 1999; Besier et al., 2003). Additionally, participants demonstrated smaller hip adduction rotation angles at IC in unanticipated turns, contrasting with findings of larger hip abduction angles in similar conditions (Norte et al., 2020; Imwalle et al., 2009). Greater hip abduction has been linked to increased knee abduction and potential ACL injury risk (Imwalle et al., 2009). Analysis of knee joint dynamics revealed notable differences during anticipated and unanticipated turns. Participants exhibited less knee flexion but increased internal rotation at peak vGRF in unanticipated turns (Besier et al., 2001). These findings align closely with previous systematic reviews indicating increased knee flexion and internal rotation angles in unanticipated single-leg cuts (Almonroeder et al., 2015). Reduced knee flexion angles can lead to greater anterior tibial shear force, a direct contributor to ACL injury (Aghdam et al., 2022; Padua and DiStefano, 2009; Favre et al., 2016; Podraza and White, 2010). Furthermore, decreased knee flexion diminishes the hamstring's ability to counteract anterior shear forces, thereby potentially increasing ACL loading (Padua and DiStefano, 2009). The ankle joint exhibited significant movement pattern changes throughout the turn phases. Notably, there was dorsiflexion at the middle and plantar flexion at the beginning and end of turns (Favre et al., 2016). Compared to anticipated turns, non-anticipatory turns showed less angular variation in sagittal and transverse sections, particularly in dorsiflexion and inversion (Podraza and White, 2010). This underscores the critical influence of anticipation on ankle stability and movement patterns during rapid changes in direction. An interaction between fatigue and anticipation was observed, particularly concerning hip angles at peak vGRF. This interaction suggests that fatigue exacerbates the challenge of predicting and executing movements, especially under unanticipated conditions. Such heightened cognitive demands may alter lower limb postures, potentially increasing the risk of ACL injuries during the braking phase of unanticipated turns (Vassalou et al., 2016). These findings have important clinical implications for injury prevention and rehabilitation strategies in athletes. The observed movement patterns and joint dynamics highlight specific vulnerabilities during unanticipated maneuvers, particularly concerning ACL injury risks associated with hip and knee joint mechanics.

Kinetic analysis

In this study, anticipatory effects on GRFs during pivot turns were also notable, with participants exhibiting higher peak vertical and posterior GRFs, alongside elevated GRF loading rates during unanticipated turns. This rapid peak loading upon ground contact indicates a reactive adaptation to unexpected movement demands. Our results align with previous research (Meinerz et al., 2015) but differ from others (Rolley et al., 2023), likely due to variations in tested movements across studies. During the braking phase of pivot turns, individuals must decelerate from sprinting speed to zero. In unanticipated scenarios, decision-making to change direction may lead to less effective coordination of body motions, resulting in a more rigid landing strategy. Increased GRFs and loading rates during rigid landings are recognized risk factors for lower limb injuries (Weinhandl and O'Connor, 2017), particularly non-contact ACL injuries (Podraza and White, 2010; Cacolice et al., 2020; Bates et al., 2013). Larger vertical GRFs increase knee loading, potentially affecting stability (Bates et al., 2013), while higher posterior GRFs may augment anterior shear forces on the tibia, potentially straining the ACL (Cacolice et al., 2020; Badiola-Zabala et al., 2020). In summary, the simultaneous alterations in knee kinetics and peak GRFs highlight an increased risk of ACL injuries during unanticipated tasks. Fatigue predominantly influenced participants' GRFs rather than joint kinematics. Post-fatigue, participants exhibited significantly lower GRFs across all components, with reduced vertical GRF loading rates. These findings are consistent with prior studies (Cortes et al., 2014; Zhang et al., 2021) reporting decreased GRFs following fatigue induction during side-step cuts and jump landings. Reduced GRFs post-fatigue may stem from diminished central nervous system excitability and muscle force production (Quammen et al., 2012; Tornero-Aguilera et al., 2022). The nature of fatigue, whether local or systemic, varied depending on protocols and participants, uniformly impairing muscle force generation and movement control, thereby diminishing overall performance. For instance, Hollman et al. (2012) found that hip extensor fatigue resulted in reduced hip flexion during jump-landing tasks among women. This underscores the nuanced effects of localized fatigue on specific joint dynamics, warranting further investigation in pivot turns.

This study revealed several key findings regarding the impact of fatigue on biomechanical parameters during pivot turns. Before fatigue induction, participants exhibited larger knee flexion moments during unanticipated turns. This observation suggests that individuals utilized knee flexors to stabilize the knee joint as the GRF vector likely passed anterior to the knee during this phase. This proactive knee flexion may serve to protect the knee from excessive loading and potential injury. Following fatigue induction, there was a slight reduction in knee flexion moment during unanticipated turns. This decline could indicate a diminished ability of knee flexors to adequately protect the knee joint under fatigue conditions. Despite this reduction in knee flexion moment, the study noted an increase in GRF during unanticipated turns post-fatigue. This discrepancy raises concerns as reduced knee flexion may compromise knee joint stability, potentially increasing susceptibility to ACL injuries. The knee joint relies on coordinated muscle actions to manage GRF effectively, and fatigue-induced alterations in muscle function could compromise this protective mechanism. Moreover, the findings underscore the complex interplay between fatigue, knee biomechanics, and injury risk during dynamic movements like pivot turns. While the study did not directly measure kinematic changes, the observed alterations in knee flexion moments highlight a critical aspect of knee joint dynamics affected by fatigue. These results align with existing literature suggesting that fatigue impairs neuromuscular control, leading to suboptimal movement patterns and increased injury risk (Fortes et al., 2024).

Limitations

This study had several limitations. Our lab-based fatigue protocol, while standardized, did not fully replicate real-match conditions. Future research should use soccer-specific tasks to better simulate sport-related fatigue. Additionally, it was difficult to separate the effects of neuromuscular and mental fatigue on performance. Future studies should isolate these effects and their interactions with anticipatory factors. Lastly, our participants were healthy soccer players. Future research should test fatigue and anticipation effects on individuals with ACL injuries and use the findings to develop targeted rehabilitation programs to prevent ACL re-tearing. Using only peak jump height to assess fatigue without considering other types of fatigue (e.g., cognitive fatigue), potentially leading to bias on the evaluation of fatigue levels. In addition, the study was conducted with uninjured soccer players. Individuals with a history of ACL injuries should be considered in the future. Finally, this study did not explore the effects of different athletic tasks on fatigue and turning performances, and the small sample size may limit the generalizability of the results.

Conclusion

In unanticipated situations, lower limbs exhibited distinct kinematic and kinetic patterns during pivot turns compared to anticipated situations. These deviations were associated with an elevated risk of ACL injury. Fatigue might impair female soccer players' limb stability production and movement control, affecting their dynamic performances. However, its effect on the risk of their ACL injury was vague. Training to enhance female soccer players' cognitive skills may potentially reduce the risk of non-contact ACL injuries in unanticipated scenarios during competitions, particularly to players rehabilitated from ACL ruptures.

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 humans were approved by the Ethics Committee of Shanghai University of Sport. The studies were conducted in accordance with the local legislation and institutional requirements. The participants provided their written informed consent to participate in this study.

Author contributions

LZ: Writing–original draft, Conceptualization. XZ: Writing–original draft. ZJ: Investigation, Writing–original draft. XW: Writing–review and editing. QZ: Writing–review and editing.

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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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Immediate changes in stroke patients' gait following the application of lower extremity elastic strap binding technique

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Objective: To ascertain the immediate changes in stroke patients' temporal and spatial parameters of gait and the joint angles of stroke patients throughout the entire gait cycle following the application of lower extremity elastic strap binding technique.

Methods: Twenty-nine stroke patients were invited as the study participants. The patient seated, flexed the hip and knee, utilized a 5 cm-wide elastic strap, positioning its midpoint beneath the affected foot and crossing it anterior to the ankle joint. Upon standing, the strap encircled the posterior aspect of the lower leg, proceeded around the back of the knee, and ascended the thigh on the affected side. Crossing anteriorly over the thigh, it then encircled the back of the waist before being secured in place. Using Qualisys motion capture system to collect kinematic data of the lower extremities during walking while wearing shoes only or strapping. A paired sample t-test was used to analyze the effects of the technique on gait spatiotemporal parameters and joint angles in stroke patients.

Results: The patients' step length decreased (P = 0.024), and step width increased (P = 0.008) during the gait cycle after the strapping. In the gait cycle between 0% and 2%, 7%–77%, and 95%–100%, the hip flexion angle on the affected side was significantly larger after the strapping ($P < 0.05$). In the gait cycle between 0% to 69% and 94%–100%, the knee flexion angle on the affected side was significantly larger after the strapping ($P < 0.05$). In the gait cycle between 0% to 57% and 67%–100%, the ankle dorsiflexion angle on the affected side was significantly smaller after the strapping ($P < 0.05$), and in the gait cycle between 0% to 35% and 68%–100%, the ankle inversion angle on the affected side was significantly smaller after the strapping ($P < 0.05$).

Conclusion: The lower extremity elastic strap binding technique can decrease the hip flexion and knee flexion limitations in stroke patients during walking, and reduce the ankle plantar flexion and ankle inversion angle of stroke patients. The lower extremity elastic strap binding technique enabled stroke patients to adopt a more stable gait pattern.

KEYWORDS

stroke, gait, lower extremity elastic strap binding technique, statistical parametric mapping, joint angles

1 Introduction

Stroke is one of the leading causes of adult death and long-term disability in China (Katan and Luft, 2018; Zhou et al., 2019). More than 50% of patients have movement disorders such as walking difficulties after a stroke, which seriously affects the quality of patients' lives (Mahlknecht et al., 2013). Reduced physical activity due to the inability to walk properly increases the risk of developing secondary health conditions (Rodríguez-Fernández et al., 2021), such as respiratory and cardiovascular complications, bowel/bladder dysfunction, obesity, osteoporosis and pressure sores, which further reduce the life expectancy of patients (Booth et al., 2012). Therefore, the restoration of independent walking ability is one of the main rehabilitation goals for stroke patients (Rodríguez-Fernández et al., 2021).

There are many traditional gait rehabilitation methods for stroke, and some new methods are also being proposed, some of which have unclear effects. Wearing ankle-foot orthoses has been proved to affect gait biomechanics (Brown et al., 2017). However, due to low comfort and high economic cost, orthoses have not been widely used in clinical practice in China (Dong et al., 2020). In recent years, scholars have designed strap binding technology which is more comfortable and lower-cost (Hwang et al., 2013). It uses elastic bandages to wrap around different parts of the patients' limb in an attempt to use the elastic characteristic of the strap to stabilize and support the various joints and the weaker muscle groups, thereby improving the abnormal gait of stroke patients. There are many ways to bind the strap, such as a single bilateral method for mild foot drop, insufficient bilateral hip extension, and poor trunk stability (Liu et al., 2022; Liu et al., 2023). A single, unilateral method for correcting ankle inversion and external rotation of the hip (Liu, 2017; Wang et al., 2018; Liu et al., 2022; Liu et al., 2023), and a combination of different methods (Wang et al., 2018), etc. As one of the single unilateral methods, the lower limb anterior rafting strap binding technique binds the joints of the lower limbs and finally fixes the strap on the shoulder, providing forward and upward assistance to the lower limbs of the hemiplegic side, while increasing the abdominal pressure of the patient and enhancing the core stability (Wang et al., 2018). In order to improve the ability to walk independently, stroke patients should relearn movement patterns by practicing walking repeatedly (Mayr et al., 2007). Thus, if rehabilitation therapy yields immediate improvements in gait for stroke patients, it could be considered as a viable long-term intervention. The lower limb anterior rafting strap binding technique is widely used in clinical practice. However, in the existing studies, there is no report on the immediate effect evaluation of the lower limb anterior rafting strap binding technique that only binds the affected lower extremity.

It has been found that strap binding techniques could improve patients' scores on the Fugl-Meyer scale, Berg scale and Barthel index (Zhang et al., 2016). However, the above scale-like assessment methods for evaluating lower limb function, balance function, and daily activity ability often need to be implemented by experienced doctors or occupational therapists, thus the results are subjective to a certain extent. Some more objective methods, such as analyzing the human kinematics parameters when the patient completes the movement, will help to further evaluate the rehabilitation effect of this technology (Xu et al., 2022), but there is relatively little research in this area. The study found that the strap binding technique increased the step length (Hwang et al., 2013), increased the step width (Zhang et al., 2016), increased the peak moment of hip and knee flexion (Wu et al., 2017), and reduced the ankle inversion angle (Xie et al., 2013) during the patient's walking. However, these studies all use discrete point analysis, usually focusing on peak indicators, which may introduce bias in data extraction. Statistical Parametric Mapping (SPM) is a statistical method used to analyze the difference between the entire curve rather than a single peak (Pataky et al., 2013), which can further quantify the influence of the lower limb anterior rafting strap binding technique on gait characteristics throughout the gait cycle.

Therefore, the objectives of this study are as follows: 1. To ascertain the immediate changes in stroke patients' temporal and spatial parameters of gait following the application of lower extremity elastic strap binding technique. 2. To evaluate the immediate changes of the lower extremity elastic strap binding technique on the joint angles of stroke patients throughout the entire gait cycle. Based on the above two objectives, we aim to provide a theoretical basis for the selection of gait rehabilitation measures for stroke patients. Based on the way of the strap binding, this study hypothesized that the lower limb anterior rafting strap binding technique may result in the following changes: 1. Decrease in step length, increase in step width, and an increase in the proportion of stance phase on the affected side, a decrease in the proportion of stance phase on the unaffected side, and a reduction in the unaffected-to-affected side stance phase ratio for stroke patients. 2. Increase in hip flexion angles, decrease in hip abduction and external rotation angles, increase in knee flexion angles, and a decrease in ankle dorsiflexion angles during walking compared with wearing shoes only.

2 Material and methods

2.1 Experimental participants

Inclusion criteria: ①The patients were diagnosed confirmed clinically by computed tomography or magnetic resonance imaging (Hyun et al., 2015); ②Functional Ambulation Category Scale (FAC) rated level 3 or above; ③Vital signs are stable and there are no other diseases that may affect gait; ④Clear awareness and the ability to understand basic commands; ⑤Signed informed consent; ⑥Age range: 40–70 years old.

Exclusion criteria: ①Patients with a previous clinical diagnosis of Parkinson's disease, fracture, myasthenia, or other diseases affecting gait; ②Patients with severe cognitive impairment who are unable to follow instructions.

G*Power 3.1.9.7 software for Windows (Heinrich-Heine-Universität Düsseldorf, Düsseldorf, Germany) was employed to conduct an a priori power analysis to determine the appropriate sample size with 80% power, effect size of 0.4 and an α error probability of 0.05. A nondirectional (two-tailed) analysis was applied. According to the aforementioned parameters, the sample size of 26 participants is needed to achieve the desired statistical power. A total of 29 stroke patients (25 males; 4 females) were invited as participants, and the specific information is shown in Table 1. This study was approved by the Ethics Committee of Beijing Sport University (approval document No. 2021080H).

TABLE 1 Baseline data of the participants.

	sample size(n)	age(y)	height (cm)	body mass (kg)	left affected limb(n)	right affected limb(n)	FAC level
males only	25	52.8 ± 9.3	168.7 ± 7.4	73.9 ± 7.1	12	13	4.2 ± 1.0
females only	4	55.3 ± 5.0	158.3 ± 6.1	66.7 ± 11.7	3	1	4.0 ± 0.8
all participants	29	53.1 ± 8.8	167.2 ± 8.0	72.9 ± 8.1	15	14	4.1 ± 0.9

Note: FAC, functional ambulation category scale.

FIGURE 1
Schematic diagram of lower limb anterior rafting strap binding technique.

2.2 Methods

2.2.1 Protocol

The participants wore uniform tights and brought their own sports shoes. A total of 19 passive reflective markers were placed bilaterally on each participant's lower extremity, including the anterior superior iliac spine, the anterior thigh, lateral and medial femur condyles, lateral and medial malleolus, tibial tuberosity, the heel, middle of the 2nd and the 3rd metatarsals. An additional marker was attached at the junction of the 4th and 5th lumbar spine vertebra (L4 and L5). Before and after the strapping, the participants walked on the ground for about 10 m at their comfortable speeds respectively. The participants walked to the end point, and the coordinates of all body surface markers were collected as a valid collection. The valid data were collected three times in the two states, respectively.

The participants were strapped by a hospital rehabilitation therapist, and the specific method refers to the approach of Wang et al., (2018) with some adjustments. The method is as follows: The patient sat and bent his or her hip and knee, took an elastic strap with a width of 5 cm, placed the middle of the strap on the bottom of the affected foot, and crossed it in front of the ankle joint. Then, the patient was asked to stand up, the strap was wrapped around the back of the shank, then around the back of the knee joint, and wound up along the thigh of the affected side. It was crossed in front of the thigh and continued to be wrapped around the back of the waist to be tied and fixed. The tension of the strap was adjusted to a comfortable level for the patient during the tying process. The specific strapping method is shown in Figure 1.

2.2.2 Data collection and reduction

The 6-lens Oqus 400 infrared motion capture system (Qualisys, Sweden, 200 Hz) was used to collect the trajectory data of body surface landmarks during a walking distance of 5–7 m from the starting point of the participants.

The raw 3-D trajectories of reflective markers were smoothed using a Butterworth low-pass filter with a cutoff frequency of 13.3 Hz to reduce random noise (Yu et al., 1999). A gait cycle was defined as the time from when the affected foot landed on the ground until it landed again, and the support phase was defined as the time from when one foot landed on the ground until that same foot was lifted off the ground. Step length and step width were calculated using the toe coordinates. Step length was defined as the distance between the unaffected side foot hitting the ground and the affected side foot hitting the ground in the anteroposterior direction, while step width was defined as the distance between the left and right directions of two consecutive heel strikes. The Euler angles method was employed to calculate the three-dimensional angle of each lower limb joint (Wu et al., 2002). The proportions of the stance phase of the unaffected and the affected sides were calculated, representing the time of the stance phase of one limb in relation to the whole gait cycle. The ratio of the unaffected-to-affected side stance phase was also calculated. The cubic spline interpolation method was used to standardize the angle of each lower extremity joint into 101 data points in one gait cycle, facilitating subsequent statistical analysis. Step length and width were standardized by dividing leg length (LL), defined as the distance from the greater trochanter of the femur to the ipsilateral lateral malleolar. All results were averaged by three valid datasets.

TABLE 2 Spatiotemporal parameters of participants' gait before and after lower extremity elastic strap binding technique.

parameters	unbundled	after binding	t value	*P*-value	*Cohen's d*
step length (LL)*	0.49 ± 0.19	0.47 ± 0.18	2.396	0.024	0.108
step width (LL)*	0.33 ± 0.08	0.34 ± 0.08	2.838	0.008	0.125
the proportion of stance phase on the unaffected side (%)	64.48 ± 7.41	64.14 ± 7.08	0.422	0.676	0.047
the proportion of stance phase on the affected side (%)	72.38 ± 9.38	73.38 ± 8.94	1.128	0.269	0.109
unaffected-to-affected side stance phase ratio	1.13 ± 0.16	1.14 ± 0.13	2.048	0.292	0.069

Note: LL, leg length, normalized relative to leg length, * indicates comparison before and after strap binding, $P < 0.05$.

2.2.3 Data analysis

The dependent variables (step length, step width, the proportion of stance phase on the unaffected side, the proportion of stance phase on the affected side, unaffected–to–affected side stance phase ratio) were compared before and after binding using a paired t-test in SPSS 21.0 software (SPSS, Chicago, IL, United States). Ensemble averages of the angles of hip flexion and extension, hip abduction and adduction, hip internal rotation and external rotation, knee flexion and extension, ankle dorsiflexion and plantar flexion, ankle inversion and eversion, ankle internal rotation and external rotation during the gait cycle before and after binding were analyzed using one-dimensional SPM paired sample t-test (Pataky et al., 2013). All SPM analyses were implemented in MATLAB R2013a (The MathWorks Inc., United States) using open-sourced code.

The significance level was set at 0.05. The effect size was represented by the Cohen's d. The values d = 0.2, d = 0.5, and d = 0.8 correspond to small, medium, and large effect sizes, respectively (Cohen, 2013).

3 Results

3.1 Spatio-temporal parameters

The results of the statistical analysis (Table 2) indicate that the standardized step length decreased, and the standardized step width increased after strap binding compared to the unbound condition. There were no statistically significant differences in the proportion of stance phase on the unaffected side, the proportion of stance phase on the affected side, and the unaffected-to-affected side stance phase ratio before and after the strap binding.

3.2 Joint angles

The flexion angle of the affected hip joint in the three stages of the gait cycle (0%–2%, 7%–77%, and 95%–100%) after strapping was larger than that before binding (Figure 2A, B). The hip abduction and adduction angles (Figure 2C, D) and the hip internal rotation and external rotation angles (Figure 2E, F) showed no significant changes before and after strapping. The knee flexion angle of the affected side in the two stages (0%–69% and 94%–100%) during the gait cycle of the patients after strapping was larger than that before strapping (Figure 2G, H). The ankle plantar flexion angle of the affected side in the 0%–57% and 67%–100% stages of the gait cycle after the strapping was smaller than that before binding (Figure 2I, J). The ankle inversion angle of the affected side in the 0%–35% and 68%–100% stages of the patients' gait cycle after strapping was smaller than that before binding (Figure 2K, L). The ankle internal rotation and external rotation angles (Figure 2M, N) showed no significant changes before and after strapping.

4 Discussion

The evaluation of gait kinematics is crucial for the diagnosis, treatment and rehabilitation of stroke (Zhang et al., 2023). In this study, infrared motion capture technology combined with SPM was used to analyze the immediate kinematic changes of lower extremity elastic strap binding technique on the gait cycle of stroke patients. The results showed that this technique decreased the standardized step length and increased the standardized step width during walking, increased hip flexion angles, knee flexion angles and decreased ankle plantar flexion angles. Thus, the experimental results partially supported the research hypothesis.

In this study, it was found that the lower extremity elastic strap binding technique could decrease the step length and increase the step width of stroke patients. Research has shown that the human body can improve dynamic stability by shortening the step length, bringing the center of mass closer to the support plane (Yang et al., 2016), and by increasing the step width to expand the support base. The lower extremity elastic strap binding technique stabilizes the joints of the lower limbs through the elastic characteristic of the strap band, limits excessive extension of the lower limb joints, decreases the step length, and increases the step width of the patients, resulting in a more stable gait (Hak et al., 2013; Yang et al., 2016). This is crucial information for rehabilitation practitioners and caregivers, as it allows them to tailor rehabilitation plans based on these changes, aiding patients in regaining their walking abilities and reducing the occurrence of accidents. However, the Cohen's d value is small, as the degree of change in step length and width is only affected to a small extent by the elastic characteristic of the strap band.

The lower extremity elastic strap binding technique can increase the hip flexion angles during walking in stroke patients. Previous studies have indicated that stroke patients have limited hip movement due to weak hip flexion strength on the affected side (De Quervain et al., 1996), significantly affecting walking ability and walking speed (Kim and Eng, 2004). In this study, the strap was crossed at the front of the affected thigh and wrapped around the back of the patient's waist for fixation. The elastic characteristic of

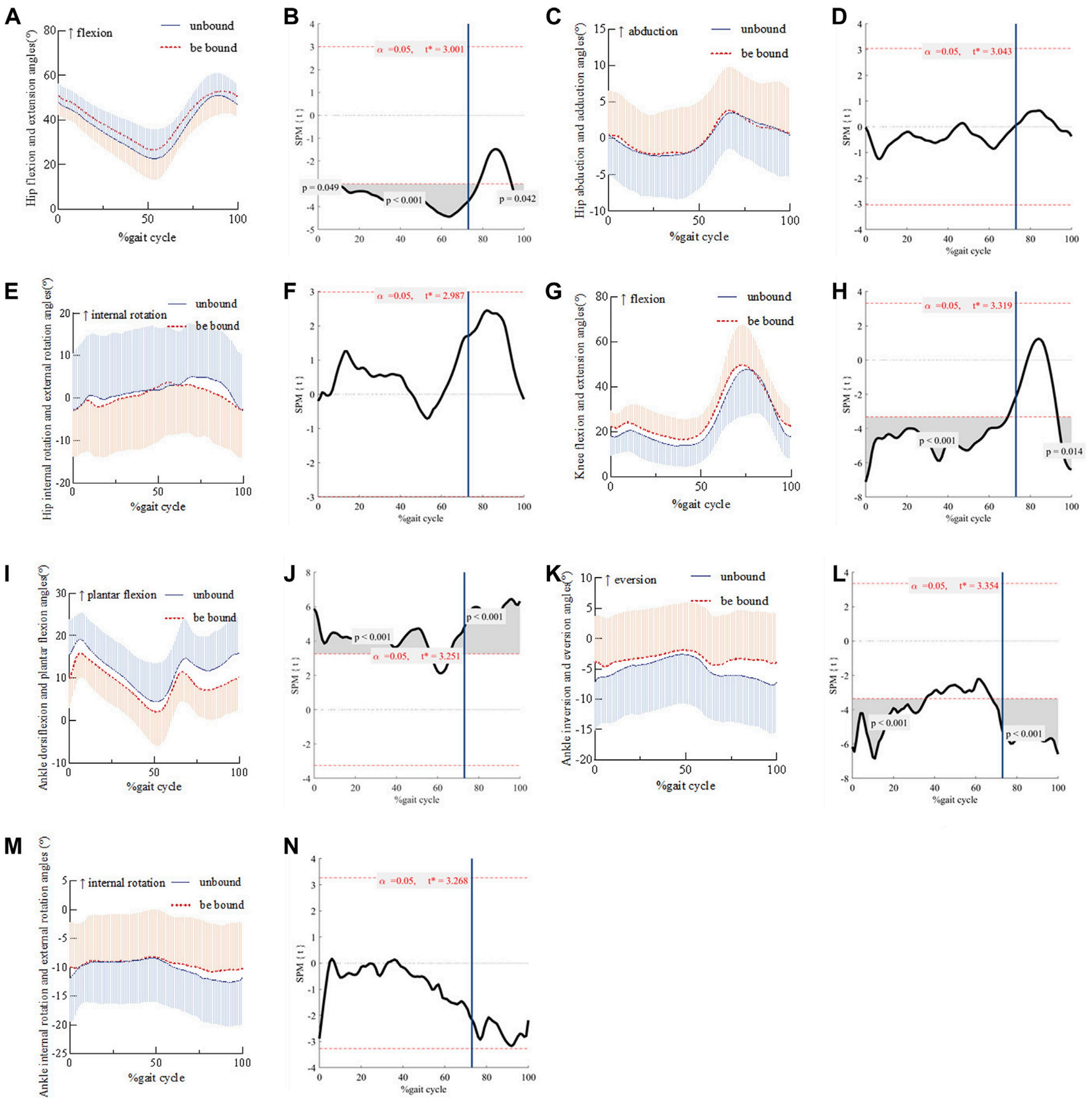

FIGURE 2
Comparison of hip flexion and extension angles and their SPM during gait cycles before and after strap binding Panel 2 **(A, C, E, G, I, K, M)** shows the change of joint angles during the gait cycle. **(B, D, F, H, J, L, N)** is the SPM(t) value, and the red dashed line indicates the critical threshold. The blue vertical line represents the moment when the affected limb leaves the ground. When the SPM(t) value exceeds the critical threshold, the difference before and after binding is statistically significant ($P < 0.05$).

the strap was used to assist hip flexion, which increased the hip flexion angle during stance phase of the patient's gait cycle.

The lower extremity elastic strap binding technique increases the knee flexion angles in stroke patients during walking. This study found that knee flexion angle increased during stance phase of the patient's gait cycle, confirming the effectiveness of wearing a strap to decrease knee hyperextension and the stiff gait pattern of the knee joint caused by muscle rigidity in stroke patients during walking. This may be attributed to the fact that the back of the affected knee joint is wrapped with the strap, providing assistance for knee bending on the affected side. Moreover, as there is a positive correlation between ankle dorsiflexion and knee flexion (Kim and Won, 2019), the resistance to plantar flexion exerted by the strap on the ankle may further increase the knee flexion angle during landing (Choo and Chang, 2021).

Anterior lower limb rafting decreases the ankle plantar flexion angles in stroke patients during walking. In stroke patients, due to weakness of the ankle dorsiflexors and evertors, or excessive activity of the plantar flexors and invertors, the foot cannot be fully lifted during the walking swing stage, sometimes accompanied by excessive inversion, which is manifested as foot drop (Prenton et al., 2016), one of the main reasons affecting gait (Choo and Chang, 2021). The elastic strap binding technique used in this study stabilizes and supports each joint through the elastic characteristic of the strap, fixes the position of the limb, and supports and stabilizes the weak muscle groups (Liu et al., 2022). It can be regarded as the combination technology of elastic band orthotics that acts on the ankle, knee, and hip. Compared with ordinary ankle-foot orthotics that restrict the patient's active dorsiflexion ability of the ankle joint and their control of movement, elastic band orthotics have less restriction on the active movement ability of the ankle joint (Xie et al., 2013). By providing appropriate plantar flexion resistance to the feet (Kim and Won, 2019), it effectively prevents the foot drop thus reduces the risk of falling in stroke patients. The strap was interlaced in front of the affected limb's ankle, and appropriate resistance to plantar flexion and inversion was applied to the patient's foot. The results showed that the plantar flexion angle and inversion angle of the ankle of the patients were reduced during most phases of the gait cycle, confirming that the technique could assist the patients in dorsiflexion and eversion of the ankle during walking, thus improving the abnormal gait.

The lower extremity elastic strap binding technique can improve the abnormal gait of stroke patients and play an auxiliary role in gait rehabilitation. Due to dysfunction, stroke patients often inhibit the use of the affected limb, resulting in insufficient use of the affected limb and loss of behavioral and neuronal function (Maier et al., 2019). Rehabilitation of stroke patients may be closely related to neural remodeling, which requires enhancing existing synaptic conduction and creating new connections (Szelenberger et al., 2020). Neuroscience studies have shown that the functional results of neuroplasticity are task-specific and depend on the nature of training (Yeung et al., 2018). This means that, in order to improve the ability to walk independently, stroke patients should relearn movement patterns by practicing walking repeatedly (Mayr et al., 2007). By making stroke patients adapt to the coupling of the walking process and being guided by the goal of normal walking, the strap binding technology produces higher activity in the sensorimotor area. It assists stroke patients in obtaining a more normal gait pattern and provides proprioceptive feedback to stimulate changes in the excitability of the motor cortex (Yeung et al., 2018). This is more conducive to the gait rehabilitation of stroke patients. Given the potential of lower extremity elastic strap binding technique to ameliorate issues of foot drop and restricted hip and knee flexion in stroke patients, in forthcoming clinical applications, rehabilitation therapists may consider employing lower extremity elastic strap binding technique for gait rehabilitation training specifically tailored for stroke patients exhibiting symptoms of foot drop and limited hip and knee flexion.

5 Limitations

In the testing of this study, a non-randomized approach was used, where all participants first underwent walking tests without binding, followed by walking tests with binding. This may introduce systematic bias. Additionally, we predicted an effect size of 0.4 for the impact of binding, which was used to calculate the sample size. However, the actual effect size was lower than 0.4, potentially leading to an underestimation of the sample size chosen, constituting one of the limitations of this study.

6 Conclusion

The lower extremity elastic strap binding technique can decrease the hip flexion and knee flexion limitations in stroke patients during walking, and reduce the ankle plantar flexion and ankle inversion angle of stroke patients. The lower extremity elastic strap binding technique can help stroke patients adopt a more stable gait pattern.

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: 10.6084/m9.figshare.25922287.

Ethics statement

The studies involving humans were approved by Ethics Committee of Beijing Sport University. The studies were conducted in accordance with the local legislation and institutional requirements. The 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: Writing–original draft, Writing–review and editing. QW: Data curation, Writing–review and editing. QL: Writing–review and editing, Data curation. XC: Data curation, Writing–review and editing. HC: Writing–review and editing, Writing–original draft. XW: Writing–original draft, Writing–review and editing.

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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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Limb dominance influences landing mechanics and neuromuscular control during drop vertical jump in patients with ACL reconstruction

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Objectives: The purpose of this study was to compare the interlimb biomechanical differences in patients who had undergone anterior cruciate ligament reconstruction (ACLR) in either dominant (ACLR-D) or nondominant (ACLR-ND) limbs and healthy controls (CON) during drop vertical jump (DVJ) task. To investigate whether the dominant or nondominant limb influences the risk of re-injury in ACLR patients.

Methods: Thirty-three ACLR patients were divided into ACLR-D and ACLR-ND groups according to whether the surgical limb was dominant or nondominant. Seventeen healthy individuals were selected as the CON group. Three-dimensional kinematic data, ground reaction force (GRF) data, and surface electromyographic (EMG) data from the bilateral lower limbs of all participants were collected during the DVJ task. Two-way repeated-measures ANOVAs (limb × group) were performed on the variables of interest to examine the main effects of limb (dominant vs. nondominant) and group (ACLR-D, ACLR-ND, and CON), as well as the interaction between limb and group.

Results: The nonsurgical limbs of ACLR group had significantly greater knee valgus angles, knee extension and valgus moments, peak posterior GRF (PPGRF), and peak vertical GRF (PVGRF) compared to the surgical limbs. The nonsurgical limbs of ACLR-ND patients demonstrated significantly greater knee extension and valgus moments, greater PPGRF and PVGRF, and reduced muscle activity in the vastus medialis and vastus lateralis compared to the CON group. The ACLR patients had reduced muscle activity in the quadriceps of the surgical limb and the hamstrings of the bilateral limbs compared to controls.

Conclusion: The nonsurgical limbs of ACLR patients may suffer an increased risk of ACL injury due to altered landing mechanics and neuromuscular control strategies compared to the surgical limbs. Additionally, limb dominance influences movement patterns and neuromuscular control during DVJ task, the nonsurgical limbs of the ACLR-ND might be at higher risk of ACL injury compared to the ACLR-D group.

KEYWORDS

ACLR, landing strategy, landing mechanics, muscle activation, return to sport

1 Introduction

Anterior cruciate ligament (ACL) injuries are among the most prevalent severe sports injuries, accounting for approximately 50% of all knee injuries (Moses et al., 2012; Kaeding et al., 2017). Following ACL injuries, patients experience abnormal neuromuscular control, decreased knee stability, and increased risk of knee osteoarthritis (Howells et al., 2011; Gersing et al., 2021). ACL reconstruction (ACLR) is a common surgical treatment following ACL injuries, contributing to restoring knee function and safely returning to play. However, a quarter of young athletic patients suffered an ACL re-injury (Wiggins et al., 2016), suggesting significantly higher injury rates compared to primary ACL injuries. The incidence rates of the surgical limb and nonsurgical limb were reported as 7%–12% and 18%–28%, respectively (Webster and Feller, 2016; Lindanger et al., 2019). Therefore, it is critical to monitor the rehabilitation progress on both surgical and nonsurgical limbs following ACLR.

Bilateral asymmetries in knee mechanics during landing were commonly observed following ACLR (Johnston et al., 2018; King et al., 2021; Kotsifaki et al., 2022b; Kotsifaki et al., 2022a), which has been considered as ACL reinjury risk factors (Johnston et al., 2018; King et al., 2021). The surgical limbs typically exhibit smaller knee flexion angles, knee extension moments, and ground reaction force (GRF) during landing (Johnston et al., 2018; Kotsifaki et al., 2022a), whereas greater knee joint contact forces and ACL forces are present in the nonsurgical limbs (Wren et al., 2018; Rush et al., 2024). In fact, the limb dominance may be associated with bilateral asymmetry in health populations during jump task (Edwards et al., 2012). Abnormal landing kinematics and kinetics for dominant and nondominant limbs, including greater valgus angles and peak GRF for nondominant limbs during jumps (Wollschläger-Tigges and Simpson, 2016; Nakahira et al., 2022), as well as higher knee extension moments and quadriceps activation for dominant limbs (Yılmaz and Kabadayı, 2022), may contribute to increased risk of ACL injuries. Therefore, whether the limb dominance contributes to the bilateral asymmetry in ACLR patients need to be investigated.

In fact, recent work reported the bilateral biomechanical characteristics in relation to the limb dominance following ACLR; however, the results for dominant and nondominant are inconclusive (Dos' Santos et al., 2019; Malafronte et al., 2021; Farmer et al., 2022; Goto et al., 2022). A recent study showed that for the surgical limb, patients underwent ACLR on the nondominant limb had greater knee loading (peak knee extension moments, peak patellofemoral joint stresses) during walking compared to patients underwent ACLR on the dominant limb (Goto et al., 2022). Conversely, Malafronte et al. (2021) reported that patients with ACLR on the dominant limb demonstrated greater knee joint loading in surgical limb compared to nondominant ACLR during jump-landing task. Meanwhile, for nonsurgical limbs, the results between dominant and nondominant limbs seem to be contradictory. Goto et al. (2022) found that dominant ACLR patients carried 49% more knee load in walking than nondominant ACLR patients. However, Malafronte et al. demonstrated that dominant ACLR patients carried 76% less knee load than nondominant ACLR patients performing jump-landing task (Malafronte et al., 2021). The above results suggest that there may be biomechanical differences between dominant and nondominant limbs in patients with ACLR, but the findings regarding the risk of secondary ACL injury or graft rupture in the surgical and nonsurgical limbs are inconsistent across studies.

The purpose of this study was to compare the biomechanical characteristics of bilateral limbs in patients who had undergone ACLR in either dominant (ACLR-D) or nondominant (ACLR-ND) limbs and healthy controls (CON) during drop vertical jump (DVJ) task. We hypothesized that (1) the nonsurgical limbs would exhibit smaller knee flexion angles, greater GRFs, greater knee extension and valgus moments, and greater quadriceps and hamstring muscle activation compared to the surgical limbs, regardless of ACLR-D or ACLR-ND group, and (2) the nonsurgical limbs in the ACLR-ND group would exhibit smaller knee flexion angles, greater knee extension and valgus moments, greater GRFs, and greater quadriceps and hamstring muscle activation compared to the nonsurgical limbs in the ACLR-D group.

2 Material and methods

2.1 Participants

Based on an estimated effect size of 0.78 for differences in knee extension moments between limbs of the ACL-D and ACL-ND (Goto et al., 2022), a sample size of 12 was required to achieve a power of 80% at a type I error rate of 0.05. A total of 50 male participants were recruited to complete this study, including three groups: (1) patients who underwent ACLR on their dominant limb (ACLR-D group, n = 17); (2) patients who underwent ACLR on their nondominant limb (ACLR-ND group, n = 16); (3) Healthy individuals matched for age, height, weight, and physical activity level to the ACLR patients, were selected as the control group (CON group, n = 17).

The patients with ACLR were recruited from Qilu Hospital of Shandong University, and the participants of CON group were recruited from Shandong Sport University. The inclusion criteria for this study were as follows: (1) aged 18–40 years; (2) Unilateral hamstring tendon reconstruction without combined meniscal medial collateral ligament injury; (3) hospital-assessed to meet criteria for return to sport; (4) 10–14 months after ACLR; (5) Willingness to return to sports (RTS) after ACLR;

TABLE 1 Participant information (mean ± SD).

	ACLR-D (n = 17)	ACLR-ND (n = 16)	CON (n = 17)	One-way ANOVA/T-test	
				F/t value	P-Value
Age (years)	24.1 ± 4.3	23.9 ± 1.7	23.4 ± 1.6	0.242[a]	0.786
Height (cm)	176.4 ± 5.1	175.9 ± 5.7	178.1 ± 6.8	0.601[a]	0.553
Weight (kg)	76.6 ± 9.4	72.7 ± 11.3	73.6 ± 15.4	0.475[a]	0.625
Dominant limb, right/left (n)	11/6	11/5	16/1	NA	NA
Postoperative duration (months)	12.1 ± 1.6	11.8 ± 1.6	NA	0.442[b]	0.662
IKDC (score)	87.2 ± 9.4	86.6 ± 6.4	NA	0.447[b]	0.658
Tegner Activity Scale (score)	6.9 ± 1.4	6.6 ± 1.4	6.8 ± 1.2	0.356[a]	0.703

IKDC, International Knee Documentation Committee; ACLR-D, anterior cruciate ligament reconstruction on dominant limb; ACLR-ND, anterior cruciate ligament reconstruction on nondominant limb; CON, control; NA, not available.
[a]F-value for one-way ANOVA.
[b]t-value for independent samples T-test.

(6) both pre-injury ACLR patients and healthy athletes regularly participated in at least one physical activity daily; (7) Tegner Activity Scale ≥5. The exclusion criteria were as follows: (1) knee-related injury within 3 months; (2) previous other knee-related surgeries; (3) severe cardiovascular and neurological disease history; (4) visual impairment and intolerable associated organ disease. The study was approved by the Ethics Committee of Sports Science of Shandong Sports University (approval number: 2023004) and registered with the China Clinical Trial Registry (registration number: ChiCTR2300076299). All patients signed the informed consent form before participation.

2.2 Procedures

This cross-sectional study design was completed in the biomechanics laboratory of the Shandong Sport University. Participants were recruited between Oct. 2023 and May 2024. Before the biomechanical assessment, participants completed the International Knee Documentation Committee (IKDC) to assess knee function. Participants' demographic information, surgery information, and dominant limb are shown in Table 1. Limb dominance was determined by which limb they were more accustomed to using when kicking a ball (Zumstein et al., 2022).

Participants changed into spandex pants and t-shirts and wore running shoes provided by the laboratory. They were allowed to perform self-selected warm-up activities for 5 min before testing. Fifty-three reflective markers were placed on the head, trunk, and limbs, with three marker clusters placed on each thigh and shank (Figure 1). Twenty electrodes were placed bilaterally on the vastus medialis (VM), vastus lateralis (VL), rectus femoris (RF), biceps femoris (BF), and semitendinosus (ST) (He et al., 2022; Di Giminiani et al., 2023).

Following a static calibration trial, participants conducted three successful trials of a DVJ task, along with three 5-s maximal voluntary isometric contraction (MVIC) tests for the quadriceps and hamstrings. For the DVJ task, participants were asked to jump forward from a 30 cm-high box onto force platforms, and immediately jump as high as possible (Baellow et al., 2020) (Figure 2). Participants landed on the two force plates with both feet respectively without falling, and all signals were collected which was considered as a successful trial. Participants were allowed to swing their arms as needed during jumps. The MVIC test for quadriceps were performed with participants in sitting with 60° of knee flexion, while hamstrings were performed with 30° of knee flexion in prone position (Kotsifaki et al., 2022b). Participants were given a 1-min rest between trials to reduce the effects of fatigue.

The three-dimensional positions of the reflective markers were captured using 12 infrared cameras at a sampling frequency of 200 Hz (Vicon Motion Systems Ltd., Oxford, United Kingdom). Bilateral ground reaction forces (GRF) data were collected using two force platforms (AMTI, Inc., Watertown, MA, United States) at a sampling frequency of 1,000 Hz. Electromyographic (EMG) signals were collected using a wireless surface EMG system (Noraxon, Arizona, United States) at a sampling frequency of 2000 Hz. The coordinate signals of markers and analog signals of GRF and EMG data collection were time synchronized using Nexus software (Vicon Motion Systems Ltd., Oxford, United Kingdom).

2.3 Data reduction

Raw marker coordinates and GRF data were filtered using a fourth-order, zero-phase Butterworth filter at a low-pass of 10 Hz (Kim et al., 2015) and 50 Hz (Teng et al., 2017), respectively. Knee joint angles were using a Cardan X-Y-Z sequence of rotations, defined as the angle between the distal and proximal segments (Kotsifaki et al., 2022b). Knee joint moments were computed using the inverse dynamics approach (Mausehund and Krosshaug, 2024). Posterior peak GRF (PPGRF) is the first peak of the posterior GRF (Dai et al., 2015), and peak vertical GRF (PVGRF) is the maximum vertical GRF in the first landing-impact phase (time

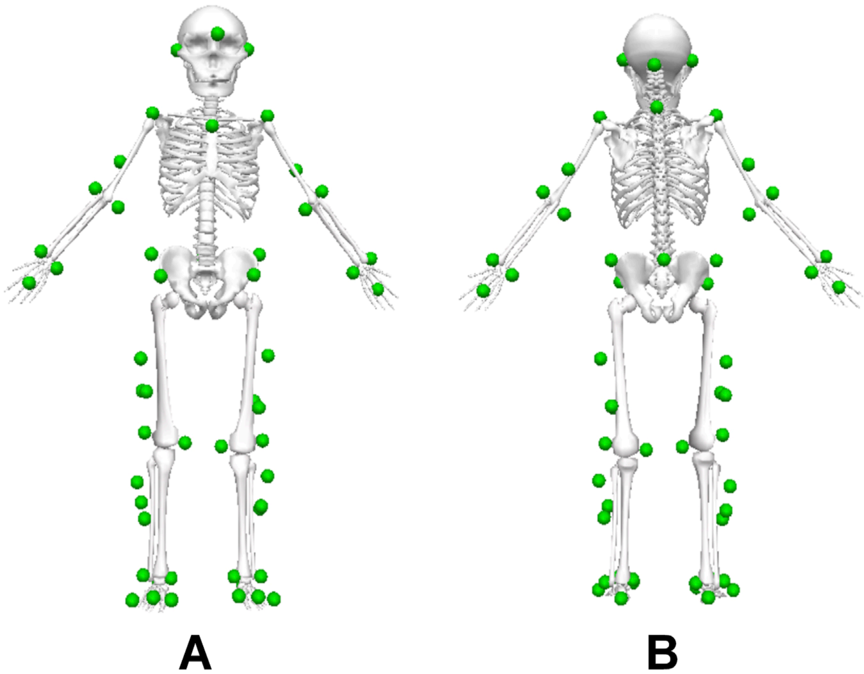

FIGURE 1
Reflective marker positions in the Visual 3D model. **(A)** view from the front. **(B)** view from behind.

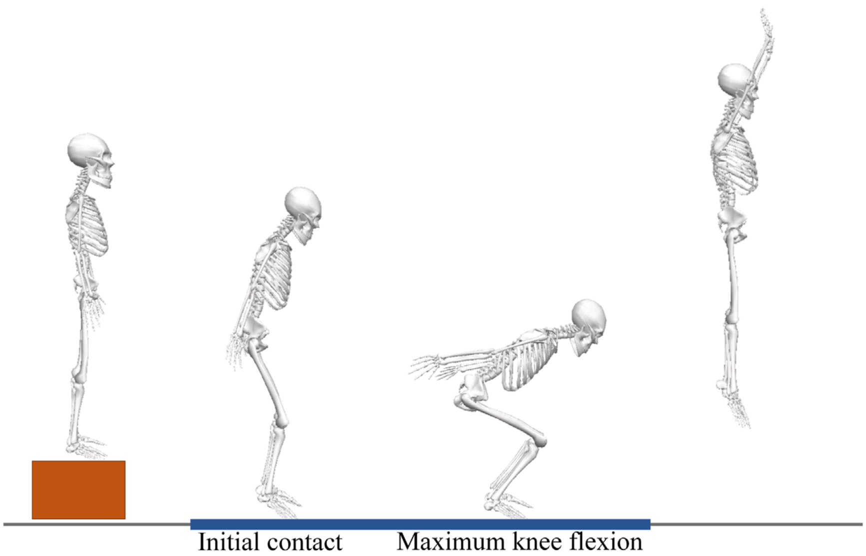

FIGURE 2
Drop vertical jump (DVJ) task. First landing-impact phase (time between initial contact with the ground and maximum knee flexion) of the DVJ task was analyzed.

between initial contact with the ground and maximum knee flexion). Posterior and vertical GRF and knee joint moments were normalized to body weight (kg).

Raw EMG signals for MVIC and dynamic tasks were filtered with a 20–500 Hz bandpass filter (Markström et al., 2022), and smoothed using a root-mean-square algorithm with a 50 milliseconds moving window (Frank et al., 2016).

The integral of EMG (IEMG) signals for assessing muscle activity during the first landing-impact phase for each muscle was calculated using the following Equation 1 (Urbanek and Van Der Smagt, 2016; Baellow et al., 2020):

$$\mathrm{IEMG} = \int_{t_1}^{t_2} |X(t)|dt \quad (1)$$

t_1 is initial contact, t_2 is maximum knee flexion, X(t) is the EMG signal. The mean time of the first landing-impact phase of the three DVJ tasks for each participant was used to calculate the IEMG in the MVIC task.

The dynamic IEMG data were normalized to the MVIC tests, therefore IEMG data were reported as %MVIC. All data processing was performed in Visual 3D software (C-Motion Inc., Germantown, United States).

2.4 Statistical analysis

Data normality was determined using the Shapiro-Wilk test. One-way ANOVAs or independent t-tests were used to compare differences in participants' demographic information among groups. Two-way repeated-measures ANOVAs (limb × group) were performed for variables of interest to examine the main effects of limb (dominant vs. nondominant) and group (ACLR-D, ACLR-ND, and CON), as well as the interaction between limb and group. Paired and independent t-tests were used to *post hoc* tests to compare differences between limbs and groups, respectively, if no significant interaction effect was detected but significant main effects were detected. One-way ANOVAs were used to determine the effects of each independent variable on a given dependent variable if a significant interaction effect was detected. The significant level was set at α = 0.05. Partial η^2 (η^2_p) was used to indicate the effect sizes of two-way ANOVAs for the main and interaction effects. The thresholds for η^2_p were: 0.01–0.06 for small, 0.06–0.14 for medium, and greater than 0.14 for large effect sizes (Pierce et al., 2004). All data were statistics in SPSS 26.0 and presented as mean ± SDs.

3 Results

Significant limb × group interactions were observed for knee valgus angle (p = 0.019; η^2_p = 0.172), knee extension moment (P < 0.001; η^2_p = 0.318), and knee valgus moment (P = 0.027; η^2_p = 0.142). *post hoc* tests demonstrated that the nonsurgical limbs of the ACLR-D and ACLR-ND groups had significantly greater knee valgus angle and knee extension moment compared to the surgical limbs, and the knee valgus moments of the nonsurgical limbs were greater than the surgical limbs in ACLR-ND group. For the nonsurgical limbs, the ACLR-ND group exhibited significantly greater knee valgus moments compared to the CON group. No any main effects or interactions were observed in knee flexion angle, knee external rotation angle, and knee internal rotation moment (Table 2).

Significant limb × group interactions were observed for the muscle activation in VM (P = 0.007; η^2_p = 0.203), RF (P = 0.007; η^2_p = 0.195), and VL (P = 0.006; η^2_p = 0.206). Post hoc tests demonstrated that the muscle activation of the surgical limbs on VM, RF, and VL in the ACLR patients and of the nonsurgical limbs on VM and VL in the ACLR-ND group was significantly lower than that in the CON group. The nonsurgical limbs of the ACLR-D group had significantly greater muscle activation in VM, RF, and VL compared to the surgical limbs (Table 3).

No significant limb × group interactions were found for the muscle activation in BF and ST, while a significant group effect was detected for both BF (P < 0.001; η^2_p = 0.398) and ST (P < 0.001;

TABLE 2 Knee joint angles and moments at PPGRF during the landing phase in drop vertical jump (DVJ) task (mean ± SD).

	ACLR-D		ACLR-ND		CON		P (η^2_p)		
	Nonsurgical	Surgical	Nonsurgical	Surgical	Dominant	Non-dominant	Limb	Group	Interaction
Knee flexion angle (°)	28.4 ± 8.6	30.1 ± 9.0	30.9 ± 8.1	31.6 ± 8.4	31.9 ± 10.0	32.4 ± 9.1	0.314 (0.022)	0.573 (0.023)	0.842 (0.007)
knee valgus angle※ (°)	−2.9 ± 2.2	−1.4 ± 0.9[a]	−3.1 ± 2.0	−1.5 ± 1.1[a]	−1.7 ± 1.1	−1.9 ± 1.0	-	-	0.019 (0.172)
knee external rotation angle※ (°)	−4.2 ± 3.0	−4.9 ± 4.7	−5.4 ± 3.9	−4.6 ± 3.6	−5.7 ± 4.6	−4.5 ± 3.8	0.519 (0.009)	0.881 (0.005)	0.485 (0.030)
knee extension moment (Nm/kg)	1.36 ± 0.46	1.19 ± 0.36[a]	1.67 ± 0.46[b]	1.21 ± 0.26[a]	1.31 ± 0.26	1.32 ± 0.31	-	-	<0.001 (0.318)
knee valgus moment※ (Nm/kg)	−0.14 ± 0.11	−0.13 ± 0.08	−0.25 ± 0.23[b]	−0.11 ± 0.07[a]	−0.11 ± 0.06	−0.12 ± 0.07	-	-	0.027 (0.142)
knee internal rotation moment (Nm/kg)	0.03 ± 0.11	0.02 ± 0.04	0.04 ± 0.09	0.02 ± 0.05	0.03 ± 0.06	0.02 ± 0.02	0.277 (0.025)	0.973 (0.001)	0.916 (0.004)

PPGRF, peak posterior ground reaction force; ACLR-D, anterior cruciate ligament reconstruction on dominant limb; ACLR-ND, anterior cruciate ligament reconstruction on nondominant limb; CON, control.
※ Knee valgus angle, knee external rotation angle and knee valgus moment were defined as negative numbers.
[a]Significant difference within-group.
[b]Significant difference compared with CON, group.

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TABLE 3 Landing-impact time and muscle activation during the landing phase in drop vertical jump (DVJ) task (mean ± SD).

	ACLR-D		ACLR-ND		CON		P (η_p^2)		
	Nonsurgical	Surgical	Nonsurgical	Surgical	Dominant	Non-dominant	Limb	Group	Interaction
landing-impact time (s)	0.278 ± 0.048	0.274 ± 0.053	0.277 ± 0.050	0.281 ± 0.055	0.294 ± 0.052	0.296 ± 0.051	0.904 (0.001)	0.510 (0.028)	0.266 (0.055)
VM (%MVIC)	113.6 ± 66.7	64.3 ± 40.1[ab]	69.9 ± 35.6[a]	63.1 ± 36.5[a]	154.8 ± 43.9	170.5 ± 66.5	-	-	0.007 (0.203)
RF (%MVIC)	84.3 ± 62.0	45.3 ± 32.8[ab]	63.3 ± 26.0	42.8 ± 29.4[a]	100.5 ± 27.4	108.9 ± 43.1	-	-	0.007 (0.195)
VL (%MVIC)	94.5 ± 52.3	53.1 ± 34.4[ab]	58.4 ± 35.9[a]	59.5 ± 42.4[a]	132.2 ± 45.3	144.8 ± 51.3	-	-	0.006 (0.206)
BF (%MVIC)	15.9 ± 9.1[a]	13.3 ± 8.7[a]	10.9 ± 4.9[a]	15.9 ± 8.6[a]	30.3 ± 18.5	25.6 ± 7.9	0.682 (0.004)	<0.001 (0.398)	0.095 (0.101)
ST (%MVIC)	13.7 ± 8.1[a]	15.8 ± 9.0[ab]	11.9 ± 4.8[a]	20.0 ± 13.0[ab]	25.5 ± 8.2	30.6 ± 12.3[b]	0.002 (0.213)	<0.001 (0.382)	0.283 (0.058)

VM, vastus medialis; RF, rectus femoris; VL, vastus lateralis; BF, biceps femoris; ST, semitendinosus; MVIC, maximal voluntary isometric contraction; ACLR-D, anterior cruciate ligament reconstruction on dominant limb; ACLR-ND, anterior cruciate ligament reconstruction on nondominant limb; CON, control.

[a]Significant difference compared with CON, group.

[b]Significant difference within-group.

η_p^2 = 0.382), as well as a main effect for limb in ST (P = 0.002; η_p^2 = 0.213). Post hoc tests demonstrated that the BF and ST activation in ACLR patients were significantly smaller compared to the CON group. Additionally, the nonsurgical limbs of the ACLR patients exhibited significantly smaller ST activation compared to the surgical limbs. No significant differences between or within groups were detected in the landing-impact time during the DVJ task (Table 3).

No significant limb × group interactions were detected on any muscle activation in MVIC tasks. Significant group main effects were observed only in muscle activation of VL (P = 0.016; η_p^2 = 0.161) and BF (P = 0.003; η_p^2 = 0.219) in the MVIC task. Post hoc tests demonstrated that ACLR patients had significantly lower activation of both VL and BF in bilateral limbs than the CON group (Table 4).

Significant limb × group interactions were observed for PPGRF (P = 0.006; η_p^2 = 0.199) and PVGRF (P = 0.029; η_p^2 = 0.140). Post hoc tests demonstrated that the nonsurgical limbs of the ACLR-D and ACLR-ND groups had significantly greater PPGRF and PVGRF compared to the surgical limbs. For the nonsurgical limbs, the ACLR-ND group exhibited significantly greater PPGRF and PVGRF compared to the CON group. Additionally, the ACLR-ND group showed greater PPGRF in the nonsurgical limbs compared to the ACLR-D group (Figure 3).

4 Discussion

The results of this study partially support our first hypothesis, indicating that the nonsurgical limbs exhibited greater GRFs and knee joint moments compared to the surgical limbs in both ACLR-D and ACLR-ND groups, with the exception of quadriceps and hamstring muscle activation. The results of this study demonstrated that the nonsurgical limbs of ACLR patients exhibited greater PPGRF and PVGRF compared to surgical limbs in the DVJ task. These findings were consistent with previous studies that have shown ACLR patients reduce the weight bearing of the surgical limbs during exercises (Song et al., 2023; Baumgart et al., 2017) due to quadriceps inhibition and weakness (Palmieri-Smith et al., 2019; Pietrosimone et al., 2022). This self-protective mechanism (Baumgart et al., 2017) reduces the impact of GRF on the surgical knee, potentially mitigating the risk of further injury. The current study results suggested a possible change in the movement pattern and neuromuscular control strategy used by ACLR patients when performing the DVJ, which may be characterized by an altered landing strategy. Despite the synchronous movement of both limbs during the DVJ, ACLR patients actively shift their center of gravity towards the nonsurgical limbs, resulting in greater GRF being absorbed by the nonsurgical limbs, which may contribute to an increased risk of ACL injury. Previous studies have shown that greater GRF can increase tibiofemoral joint compression forces, which is a known risk factor for ACL injury (Meyer et al., 2008; Boden et al., 2010). Therefore, these findings suggested that patients in both the ACLR-D and ACLR-ND groups may be at an increased risk of ACL injury in their nonsurgical limbs compared to the surgical limbs during DVJ task, potentially due to the altered movement patterns and neuromuscular control strategies.

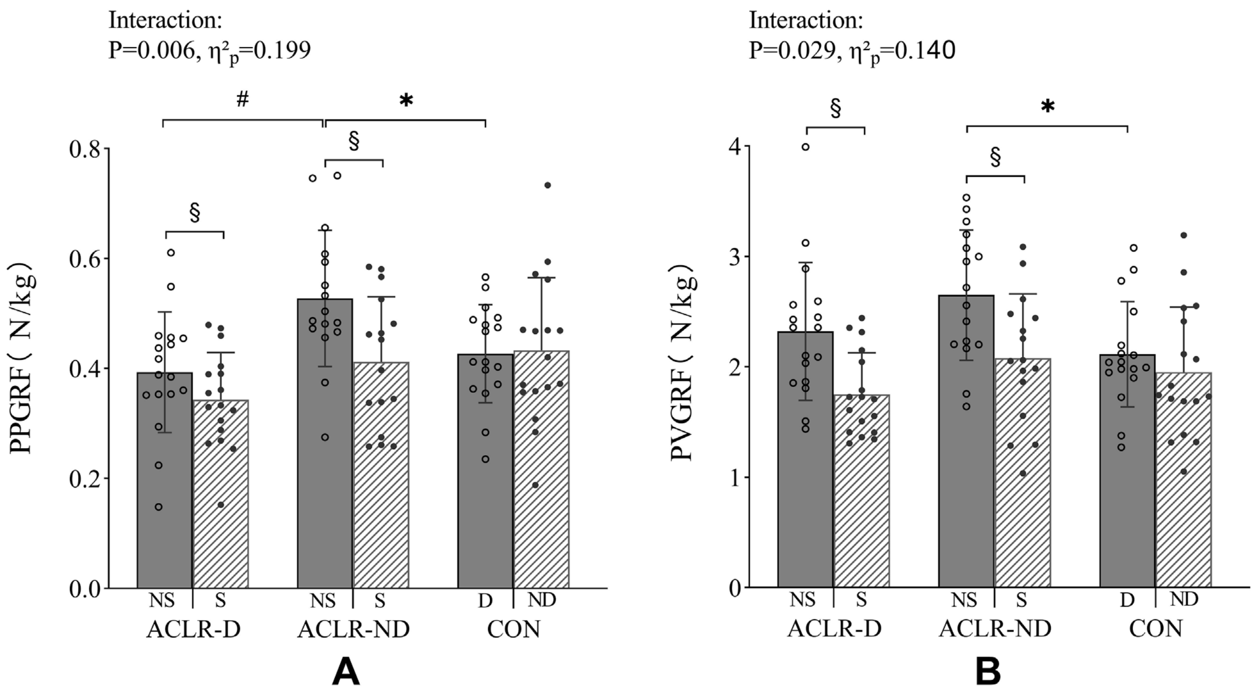

FIGURE 3
PPGRF **(A)** and PVGRF **(B)** in the first landing-impact phase of the DVJ task. PPGRF, peak posterior ground reaction force. PVGRF, peak vertical ground reaction force. NS, nonsurgical limb. S, surgical limb. D, dominant limb. ND, nondominant limb. ACLR-D, anterior cruciate ligament reconstruction on dominant limb. ACLR-ND, anterior cruciate ligament reconstruction on nondominant limb. CON, control. * Significant difference compared with CON group. # Significant difference compared with ACLR-ND group. § Significant difference within-group.

The current study revealed no significant differences in knee angles and moments at PPGRF between the dominant and nondominant limbs of healthy individuals during the DVJ task. This may suggest that ACLR patients have no inherent differences in the bilateral limbs prior to the ACL injury. Conversely, our results indicated that ACLR patients demonstrated greater knee valgus angles and extension moments in their nonsurgical limbs compared to their surgical limbs, as well as the ACLR-ND patients also exhibited greater knee valgus moments in their nonsurgical limbs. This is consistent with previous studies that have reported reduced surgical knee loading in patients with ACLR (Sritharan et al., 2020; Kotsifaki et al., 2022b; Bühl et al., 2023). The reason for these results may be an adaptive change in the landing strategy of ACLR patients. ACLR patients may rely more heavily on their nonsurgical limbs due to decreased VM, VL and RF muscle strength, impaired knee proprioception, and reduced stability in the surgical limbs (Howells et al., 2011; Arumugam et al., 2021), which can lead to an adaptive change in their landing strategy. This also validated that asymmetry of knee moments is associated with asymmetric GRF, which can lead to altered movement patterns and increased risk of injury (Dai et al., 2014). As results of these adaptive changes in landing strategy, ACLR patients may be at an increased risk of ACL injury in their nonsurgical limbs, particularly during dynamic movements that involve landing.

The results of this study partially support our second hypothesis that the nonsurgical limbs in the ACLR-ND patients exhibited greater knee extension and valgus moments, as well as greater GRFs compared to CON group, which may contribute to an increased risk of ACL injury. Furthermore, in the nonsurgical limbs, the ACLR-ND patients demonstrated greater PPGRF compared to the ACLR-D patients, as well as greater GRFs, and greater knee valgus and extension moments compared to the CON group. These differences did not exist between the ACLR-D and CON groups. These were similar to previous studies on single-leg jump (Mohammadi et al., 2013), side cut (Warathanagasame et al., 2023), and stair walking (Zabala et al., 2013) tasks. However, in contrast with our results, neither Rush et al. (2024) nor Chen et al. (2024) observed greater GRF and knee extension and valgus moments in the nonsurgical limbs compared to the healthy individuals in single-leg jump or DVJ tasks. The inconsistent results may be attributed to the fact that these studies did not account for the potential influence of limb dominance on movement patterns after ACLR. Notably, the nonsurgical limb was the dominant limb in the ACLR-ND patients in the current study, which may have influenced their movement patterns and neuromuscular control strategies during the landing task. Compared to ACLR-D, the ACLR-ND patients may have been more inclined to use a protective pattern, characterized by increased knee extension and valgus moments, on the nonsurgical limbs during the landing phase, and felt more confident with aggressive landings. However, for ACLR-ND patients, this protective movement pattern may have unintended consequences, as the increased knee loading on the nonsurgical limbs may actually increase the risk of ACL injury, rather than reducing it.

Contrary to our initial hypothesis, no significant differences in muscle activation levels of the quadriceps and hamstrings were observed in the nonsurgical limbs between ACLR-ND and ACLR-D patients. However, our study revealed that muscle activation in the VM and VL of the nonsurgical limbs was significantly decreased during DVJ task

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TABLE 4 Muscle activation during the maximal voluntary isometric contraction (MVIC) task (mean ± SD).

	ACLR-D		ACLR-ND		CON		P (η_p^2)		
	Nonsurgical	Surgical	Nonsurgical	Surgical	Dominant	Non-dominant	Limb	Group	Interaction
VM_MVIC (μv·s)	54.0 ± 12.4	52.2 ± 11.5	60.0 ± 14.8	53.0 ± 19.4	66.3 ± 20.6	63.3 ± 25.4	0.106 (0.055)	0.101 (0.093)	0.657 (0.018)
RF_MVIC (μv·s)	62.4 ± 9.3	60.2 ± 9.1	67.1 ± 18.5	63.1 ± 15.4	68.5 ± 14.9	67.2 ± 16.1	0.189 (0.036)	0.331 (0.046)	0.852 (0.007)
VL_MVIC (μv·s)	55.3 ± 17.6[a]	57.8 ± 16.0[a]	57.4 ± 14.1[a]	55.9 ± 8.0[a]	67.7 ± 8.7	66.3 ± 8.7	0.946 (0.001)	0.016 (0.161)	0.475 (0.031)
BF_MVIC (μv·s)	65.4 ± 9.6[a]	64.9 ± 13.8[a]	66.8 ± 6.2[a]	68.4 ± 8.1[a]	78.8 ± 21.5	79.8 ± 14.1	0.659 (0.004)	0.003 (0.219)	0.854 (0.007)
ST_MVIC (μv·s)	78.3 ± 13.6	80.0 ± 12.2	79.9 ± 16.0	77.4 ± 8.7	83.3 ± 15.0	81.3 ± 16.4	0.687 (0.003)	0.601 (0.021)	0.714 (0.014)

VM, vastus medialis; RF, rectus femoris; VL, vastus lateralis; BF, biceps femoris; ST, semitendinosus; MVIC, maximal voluntary isometric contraction; ACLR-D, anterior cruciate ligament reconstruction on dominant limb; ACLR-ND, anterior cruciate ligament reconstruction on nondominant limb; CON, control.
[a]Significant difference compared with CON group.

in ACLR-ND patients compared to CON group, in addition to reduced activation of the quadriceps and hamstrings of the surgical limbs. Additionally, muscle activation in BF and ST of the nonsurgical limbs was significantly lower during DVJ task in both ACLR-ND and ACLR-D patients compared to CON group. These results were consistent with literatures, which also reported lower bilateral quadriceps and hamstring activation in ACLR compared to healthy controls (Alanazi et al., 2020; Einarsson et al., 2021). This may be attributed to reduced quadriceps and hamstring muscle strength and neuromuscular inhibition (Palmieri-Smith et al., 2019; Pietrosimone et al., 2022). Additionally, a recent study reported that increased quadriceps and hamstring activation was associated with reduced knee flexion angle (Malfait et al., 2016). Therefore, ACLR patients may attempt to obtain a greater knee flexion angle to reduce impact of GRF by reducing bilateral muscle activation. Lower quadriceps and hamstring activation was associated with reduced dynamic knee stability, which may be a contributing factor to the increased risk of ACL injury (Ortiz et al., 2014; Palmieri-Smith et al., 2019; Wang et al., 2023). These results combined together suggest that abnormal quadriceps and hamstring activation in ACLR-ND patients is associated with an increased risk of ACL injury. In summary, our study revealed significant differences in quadriceps and hamstring activation levels between dominant and nondominant limbs in ACLR patients. These findings emphasized the need for personalized rehabilitation programs that take into account limb dominance to optimize outcomes and reduce the risk of further injury in ACLR patients.

There are several limitations in our study. Firstly, all participants were male, and since gender differences in knee valgus angles, and GRFs during landing (Seymore et al., 2019; Peebles et al., 2020) may affect the applicability of our results to females, future studies should investigate the effects of limb dominance on biomechanics in female ACLR patients. Secondly, we only analyzed ACLR patients with autologous hamstring grafts. Since there is an effect of different graft types on knee biomechanics (Wang et al., 2018; Yang et al., 2020), further studies in patients with other graft types are needed. Third, we did not collect muscle strength, proprioception from the participants. Previous studies have indicated that muscle strength, proprioception affect knee function and athletic performance (Ma et al., 2022; Chang et al., 2024). Future studies should investigate the effect of limb dominance on functional outcomes. Fourth, we only investigated biomechanical characteristics in the DVJ task, and future studies should investigate the effects of limb dominance on biomechanics during various movement tasks, including single-leg jumps and side-cutting maneuvers. Fifth, we conducted a cross-sectional analysis, which did not allow us to examine the longitudinal effects of limb dominance on biomechanics after ACLR. The long-term effects of limb dominance on biomechanics after ACLR remain unclear and warrant further investigation.

5 Conclusion

The nonsurgical limbs of ACLR patients may suffer an increased risk of ACL injury due to altered landing mechanics and neuromuscular control strategies compared to the surgical limbs. Additionally, limb dominance influences movement patterns and neuromuscular control during DVJ task, the nonsurgical limbs of the ACLR-ND might be at higher risk of ACL injury compared to the ACLR-D group. Given that limb dominance affects movement patterns, the impact of limb dominance should be considered in the rehabilitation of ACLR patients for better return to sport.

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 study was approved by the Ethics Committee of Sports Science of Shandong Sports University (approval number: 2023004) and registered with the China Clinical Trial Registry (registration number: ChiCTR2300076299). The studies were conducted in accordance with the local legislation and institutional requirements. The 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

BX: Writing–review and editing, Writing–original draft, Visualization, Validation, Software, Project administration, Methodology, Investigation, Formal Analysis, Data curation, Conceptualization. XY: Writing–review and editing, Validation, Supervision, Methodology, Investigation. XW: Writing–review and editing, Supervision, Project administration, Methodology. CY: Methodology, Funding acquisition, Writing–review and editing. ZZ: Supervision, Resources, Project administration, Writing–review and editing, Methodology.

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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 dual-task paradigm on the injury potential during landing among individuals with chronic ankle instability

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Purpose: Chronic ankle instability (CAI) causes maladaptive neuroplastic changes in the central nervous system, which may lead to high injury potential under dual-task conditions. This study aims to explore the effects of dual-task paradigm on the injury potential during landing among individuals with CAI.

Methods: Twenty participants with CAI (4 female and 16 male, 12 were affected with their right limbs and 8 were affected with their left limbs, 20.4 ± 1.7 years, 176.9 ± 5.0 cm, and 72.0 ± 11.1 kg) and eighteen without CAI (6 female and 12 male, 20.2 ± 1.5 years, 173.5 ± 7.0 cm, and 70.3 ± 10.8 kg) were recruited. They drop-landed on a trap-door device, with their affected or matched limbs on a flippable platform, under single- (drop-landing only) and dual-task (drop-landing while subtracting of serial threes) conditions. A twelve-camera motion capture system was used to capture the kinematic data. Two-way ANOVA with mixed design (CAI vs non-CAI groups by single-vs dual-task conditions) was used to analyze the data.

Results: Significant group-by-condition interactions were detected in the ankle inversion angle (P = 0.040, η^2_p = 0.012) and ankle inversion angular velocity (P = 0.038, η^2_p = 0.114). Both indicators decreased among individuals without CAI from single-to dual-task conditions, while remained unchanged among those with CAI; and they were higher among individuals with CAI under both single- and dual-task conditions, compared to those without CAI.

Conclusion: Individuals with CAI have a reduced ability to limit ankle inversion compared to those without CAI. Under dual-task conditions, individuals without CAI limited their ankle inversion, while those with CAI did not. Drop-landing, especially under dual-task conditions, poses a high risk of excessive ankle inversion for individuals with CAI.

KEYWORDS

ankle sprain, dual-task paradigm, drop-landing, ankle inversion angle, ankle inversion angular velocity

1 Introduction

Ankle sprains are one of the most common sports injuries, accounting for 10%–20% of all sports-related injuries (Fong et al., 2007) and up to 70%–80% among college students with sports experience (Doherty et al., 2014). In the Netherlands, approximately 440,000 ankle sprains occur annually (Kemler et al., 2015), while in the United Kingdom, they constitute 3%–5% of all emergency department visits, resulting in 1–1.5 million cases per year (Gribble et al., 2016b). Acute ankle sprains can lead to repeated injuries and the development of chronic ankle instability (CAI), with a prevalence of about 20%–30% (Konradsen et al., 2002; Herzog et al., 2019). CAI is characterized by pain, instability, recurrent injuries, and persistent dysfunction (Gribble, 2019), significantly impacting an individual's physical activity levels (Hubbard-Turner and Turner, 2015).

Landing from a height is a common scenario leading to ankle sprains (Doherty et al., 2014; Ardakani et al., 2019). During landing, the foot and ankle complex absorb the impact force from the ground, which can be 2–5 times the body weight (Xu et al., 2024b; Xu et al., 2023). This force is primarily transmitted through the medial aspect of the ankle, may cause sudden and substantial inversion of the ankle joint, which can lead to ankle sprains (Koshino et al., 2017). Larger ankle inversion angles and angular velocities during landing are associated with an increased potential for ankle sprains (Xu et al., 2024a; Zhang et al., 2024). Individuals with CAI have greater ankle inversion angles and angular velocities during landing compared to those without CAI (Simpson et al., 2022; Terrier et al., 2014). Excessive ankle inversion increases the distance between the talus and fibula, stretching the ligaments connecting these bones (Fong et al., 2012). When the ligament is stretched beyond its maximum bearing capacity, it can lead to ligament tears (Medina McKeon and Hoch, 2019).

CAI, a neurophysiological disorder characterized by maladaptive neuroplastic changes in the central nervous system (CNS), may increase potential injury under dual-task conditions. Individuals with CAI often exhibit reduced activation of the dorsal anterior cingulate cortex (Shen et al., 2022), a critical region for integrating cognitive resources (Bush et al., 2000). This reduction in cognitive resources can impair motor performance and elevate injury risk in environments requiring additional cognitive demands, such as dual-task conditions (Rosen et al., 2021; Choi et al., 2023). For example, a volleyball player with CAI aiming to execute a powerful smash must jump as high as possible and extend their upper limb. This constitutes a motor task. Simultaneously, they must consider the optimal landing spot for the ball based on the opposing defender's position and condition, representing a cognitive task. Together, these actions embody a dual-task condition. Individuals with CAI, who have a limited total capacity of potential cognitive resources, experience reduced allocation of these resources to both cognitive and motor tasks. Consequently, their performance on either task may suffer, potentially leading to tactical errors or unintentional injuries.

There is ongoing controversy regarding the performance of individuals with and without CAI during dual-task conditions, as well as the potential impact of dual-task paradigms on injury risk in individuals with CAI. Some researchers report an increase in injury potential among individuals with FAI when transitioning from single-task to dual-task conditions, attributing this to impaired feedforward and feedback mechanisms of motor control within the CNS (Tavakoli et al., 2016). Conversely, others have observed enhanced postural stability among CAI individuals under dual-task conditions, which they attribute to increased conscious control over body movement due to fear of anticipated pain or reinjury (Shiravi et al., 2017). Moreover, it remains uncertain whether CAI individuals are more susceptible to the effects of dual-task paradigms compared to those without CAI. Some studies have indicated that athletes with FAI exhibit poorer postural stability than healthy controls under dual-task conditions, potentially increasing their injury risk (Rahnama et al., 2010). However, other studies have reported no differences in postural stability between CAI and non-CAI individuals under dual-task conditions (Choi et al., 2023).

Investigating the effects of dual-task paradigm on the injury potential among individuals with CAI may enhancing our understanding on whether the maladaptation of CNS affect the ankle sprain recurrence, and even the tactical arrangements of when and how long the players with CAI would be in the game. Therefore, we hypothesized that: 1) compared to individuals without CAI, those with CAI have higher injury potential, reflected by greater ankle inversion angle and angular velocity. 2) compared to single-task conditions, the injury potential increased among individuals with and without CAI under dual-task conditions.

2 Methods

2.1 Participants

Sample size calculations were conducted using G*Power 3.1 software (University of Düsseldorf, Düsseldorf, Germany). Prior to the formal study, a pilot study was performed with six participants with CAI and another six participants without CAI. The ankle inversion angle and ankle inversion angular velocity were used as outcome measures to estimate the sample size. The effect sizes ($\eta^2{}_p$) for the group comparison (CAI vs. non-CAI) by condition (single-task vs. dual-task) were 0.064 and 0.087, respectively. Based on these calculations, a total of at least 34 participants (17 in each group) were required to achieve a statistical significance level of 0.05 and a statistical power of 0.80.

Participants were recruited in a local university from April to June 2023 through distributing posters and leaflets. Following the guidelines of the International Ankle Consortium (Gribble et al., 2016a), the inclusion criteria for participants with CAI were: 1) at least one severe ankle sprain a year prior to the recruitment, causing pain, swelling, and other inflammatory symptoms, inhibiting normal participation in daily activities for more than 1 day; 2) aged 18–24 years (Ardakani et al., 2019); 3) at least two episodes of ankle "giving way" in the past 6 months; 4) persistent sense of ankle instability and impaired ability of daily activities and 5) with a score <24 of the Cumberland Ankle Instability Tool (CAIT) (Hiller et al., 2006). Inclusion criteria for participants without CAI were: 1) no previous ankle sprain/injury and no episodes of ankle "giving way" or feeling of ankle instability and 2) CAIT score >28. Exclusion criteria for all participants were: 1) self-reported history of lower limb fractures or surgery; 2) experienced acute injuries such as sprains in the lower limbs 3 months prior to the

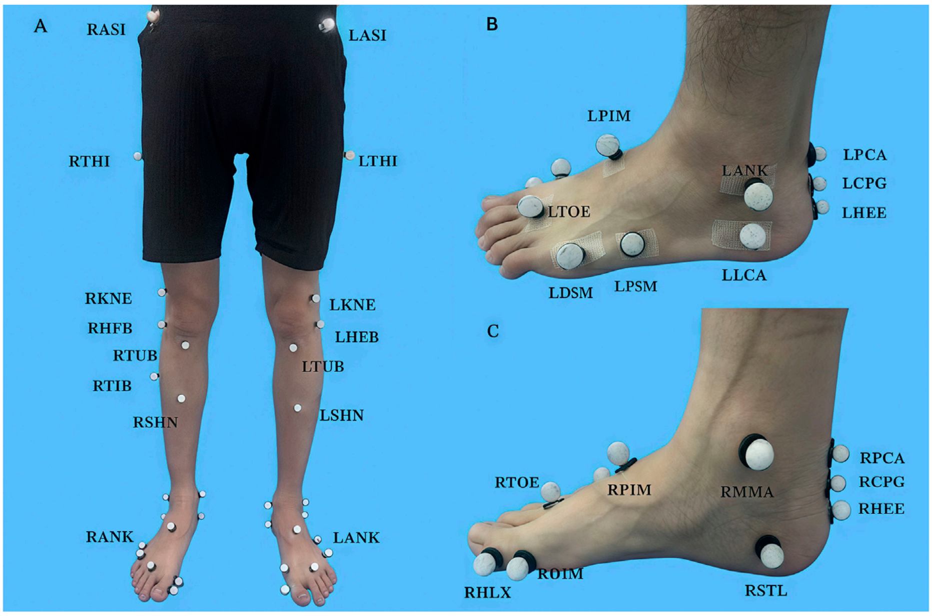

FIGURE 1
Oxford Foot Ankle Model **(A)** Front view of full markers. **(B)** Lateral view of foot markers. **(C)** Medial view of foot markers.

recruitment; 3) bilateral chronic ankle instability. All participants in this study were regular college students, both with and without CAI, who attended physical education classes three times a week for 45 min per session. They were not sedentary and were in good physical condition. After the assessment, 38 participants met the inclusion criteria, of whom 20 with CAI (4 female and 16 male, 12 were affected with their right limbs and 8 were affected with their left limbs, 20.4 ± 1.7 years, 176.9 ± 5.0 cm, and 72.0 ± 11.1 kg) and 18 without CAI (6 female and 12 male, 20.2 ± 1.5 years, 173.5 ± 7.0 cm, and 70.3 ± 10.8 kg) were recruited. This study was approved by the Institutional Review Board of Shandong Sport University (Approval Number: 2023014) and was in accordance with the Declaration of Helsinki, and all participants signed informed consent forms.

2.2 Protocols

Participants wore uniformed tight shorts and T-shirts, and 36 markers were adhered to their lower limbs, following the protocol of the Oxford Foot Ankle Model (McCahill et al., 2008) (Figure 1). The test limb in the CAI group were the affected limb, and the test limb in the non-CAI group were matched based on the ratio of the right and left affected limb in the CAI group, from which the number of right limbs tested in the non-CAI group was calculated to be: 12/20*18 = 10.8. i.e., 11 participants in the non-CAI group tested the right side, and 7 participants tested the left side. The non-CAI participants were randomized to the computerized array, which was used to determine the side of limb to be tested. Before formal tests, participants familiarized themselves with the procedure by conducting at least 3 drop-landing trials. Then, they conducted formal drop-landing tests under single- and dual-task conditions in a randomized order.

2.2.1 Drop-landing tests

Participants drop-landed from a height of 30 cm (Mackala et al., 2020) to a custom-made trap-door device consisting of three platforms, namely, take-off, flippable, and supporting platforms (Figure 2A). The height of 30 cm for landing has been proven safe and is widely applied in the literature (Shibata et al., 2023; Mokhtarzadeh et al., 2017; Watanabe et al., 2022; Li et al., 2018; Huang et al., 2024; Lim et al., 2020). The surface of the flippable platform would be flipped when subjected to a force >10 N. A marker was attached to the lateral edge of the flippable platform surface to identify the time point when it flipped. During the drop-landing test, the participants' kinematic data were recorded by a 12-camera, 3D infrared motion capture system (Vicon Vantage V5, Oxford Metrics Limited, Oxford, United Kingdom), with a frequency of 100 Hz.

Single-task conditions: Participants stood on the take-off platform, with their eyes looking straight ahead and hands on hips, and extended their affected or matched feet forward. They moved their bodies forward away from the take-off platform to minimize upward movement and then landed on the flippable and supporting platform with the affected or matched and the contralateral limbs, respectively (Figure 2). Participants performed three successful trials, defined as participants being able to stabilize their body and maintain the body position for at least 3 seconds after landing.

Dual-task conditions: During drop-landing, a subtraction of serial threes from a given three-digit number was performed simultaneously. In each trial, participants subtracted three for three times. Two subtractions were done before drop-landing, and after the second result was given, each participant extended their affected or matched feet forward and performed drop-landing as in single-task conditions, and they gave the result of the third subtraction

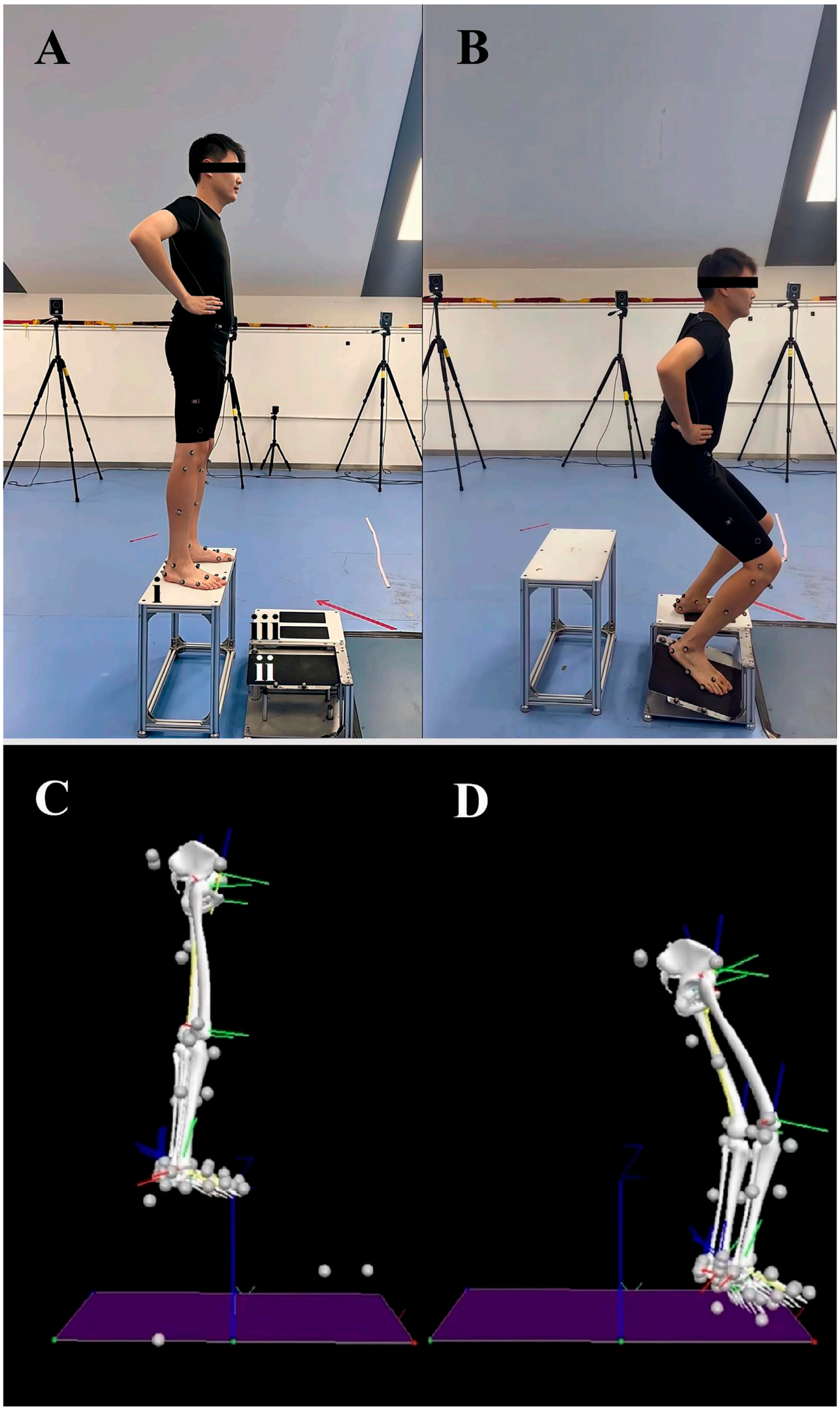

FIGURE 2
Illustration of drop-landing tests **(A)** Body position before drop-landing. i, take-off platform; ii. flippable platform; iii, supporting platform; **(B)** Body position after drop-landing; **(C)** A free body diagram of body position before drop-landing; **(D)** A free body diagram of body position after drop-landing.

immediately after landing (Figure 3). To ensure that the subtractions were calculated during landing, the timing of the third subtraction was strictly limited. Prior to the drop-landing test, participants performed consecutive subtractions for three times while sitting in a chair, and the mean time of the three subtractions were recorded as the "sitting subtraction time" (Wrightson et al., 2020). If the

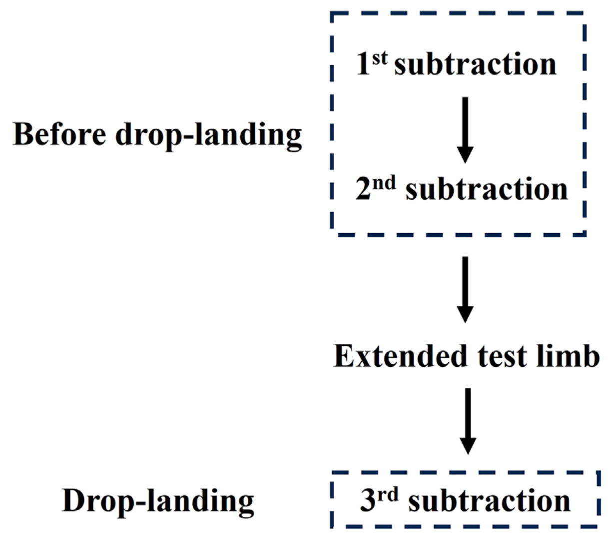

FIGURE 3
Illustration of drop-landing test under dual-task conditions.

time taken for the third subtraction under dual-task conditions during drop-landing exceeded the "sit-subtraction time", the trial would be deemed unsuccessful. Participants performed three successful trials.

2.3 Data reduction

Kinematic data were imported into Visual 3D (V6 professional, C-Motion, Maryland, United States), and low-pass filtered with a cutoff frequency of 12 Hz (Ford et al., 2007). Based on the marker protocol, an 11-segment Oxford foot ankle model was created and embedded in the motion capture files. The data were collected from the time point at landing to 200 ms after landing, defined by the movement of the markers affixed to the lateral edge of the flippable platform surface (Fu et al., 2014). The landing period was chosen because real ankle sprain occurs within it (Fong et al., 2009). The axis of ankle inversion/eversion is defined as the floating axis, which is the common axis perpendicular to both the Z-axis in the tibia/fibula coordinate system (the line extending from the tip of the medial malleolus to the tip of the lateral malleolus, directed rightward) and the Y-axis in the calcaneus coordinate system (the line aligning with the longitudinal axis of the tibia/fibula in the neutral position, pointing cranially) (Wu et al., 2002).

2.4 Variables

The ankle inversion angle was defined as the angle of rotation of the foot coordinate system relative to the tibial coordinate system in the coronal plane during the landing period.

The ankle inversion angular velocity is defined as the peak rate of the change of ankle inversion angle during the landing period, i.e., the peak value of angle increment per unit time.

2.5 Statistics

The normality of data distribution was examined using Shapiro-Wilk tests. Mixed model two-way ANOVAs were utilized to compare ankle inversion angle and angular velocity between single- and dual-task conditions among participants with and without CAI. If significant group (CAI vs non-CAI) by condition (single-task vs dual-task) interactions were detected, stratified t-tests with Bonferroni adjustment were used to conduct pairwise comparisons. Partial eta squared (η^2_p) was used to indicate the effect size of the two-way ANOVA's interactions and main effects with the thresholds: 0.01~0.06 for small, 0.06~0.14 for moderate, and > 0.14 for large effect size (Pierce et al., 2004). Cohen's d was used to indicate the effect size of *post hoc* pairwise comparison with the thresholds: <0.20 for trivial, 0.21~0.50 for small, 0.51~0.80 for medium, and >0.81 for large effect size (Cohen et al., 1988). The significance level was set at 0.05.

3 Results

The Shapiro-Wilk test indicated that all the dependent variables were normally distributed (P > 0.05). Significant group-by-condition interactions were detected in the ankle inversion angle (P = 0.04, η^2_p = 0.012) and ankle inversion angular velocity (P = 0.038, η^2_p = 0.114). *Post hoc* analysis showed that compared to single-task conditions, the ankle inversion angle (P = 0.003, d = 0.84) and ankle inversion angular velocity (P = 0.007, d = 0.91) were significantly decreased among individuals without CAI under dual-task conditions, whereas no significant differences were detected among individuals with CAI. Compared to those without CAI, the individuals with CAI had greater inversion angles under single- (P = 0.045, d = 0.65) and dual-task (P < 0.001, d = 1.13) conditions, and similarly, they had greater ankle inversion angular velocities under single- (P = 0.164, d = 0.46) and dual-task (P = 0.001, d = 1.03) conditions (Figure 4).

4 Discussion

To our knowledge, this is the first study to determine the effects of the dual-task paradigm on the injury potential during drop-landing among individuals with and without CAI. The results supported Hypothesis 1, which predicted that individuals with CAI would have a higher injury potential compared to those without CAI, under both single-task and dual-task conditions. This was reflected by a greater ankle inversion angle and angular velocity among individuals with CAI. Conversely, Hypotheses 2 was not supported by the results. Hypothesis 2 proposed that, compared to single-task conditions, the injury potential would be higher among individuals with and without CAI under dual-task conditions, but this was not observed.

The results indicated that individuals with CAI exhibited greater ankle inversion angle and angular velocity compared to those without CAI under both conditions. This finding is supported by a previous study, which showed that individuals with CAI exhibited greater ankle inversion angle after initial contact during landing compared to copers (individuals who suffered ankle sprain but

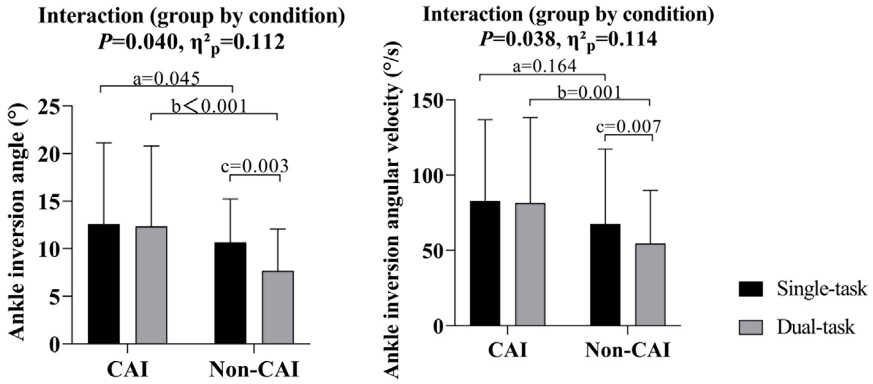

FIGURE 4
Ankle inversion angle and angular velocity in the CAI and non-CAI populations under single- and dual-task conditions. a. Significant difference between CAI and non-CAI populations under single-task conditions. b. Significant difference between CAI and non-CAI populations under dual-task conditions. c. Significant difference between single- and dual-task conditions in non-CAI populations.

did not develop CAI) and healthy controls (Oh et al., 2024). We propose that the inability to limit excessive ankle inversion during landing is linked to dysfunction in ankle eversion muscles and lateral ankle ligaments. The ankle eversion muscles, especially the peroneal muscles, have been proved to have prolonged reaction time during sudden ankle inversion (Menacho Mde et al., 2010), as well as difficulty generating maximum ankle eversion torque in an inverted position among individuals with CAI (Dong et al., 2024). In the case of sudden ankle inversion, the injured anterior talofibular ligament is unable to limit the separation between the talus and fibula, making it difficult to minimize the ankle inversion angle and angular velocity (Rigby et al., 2015). As a result, individuals with CAI are more likely to be injured than individuals without CAI under both single- and dual-task conditions.

Contrary to Hypothesis 2, a surprising finding was that compared to single-task conditions, ankle inversion angle and angular velocity were lower under dual-task conditions among individuals without CAI. This may because participants adopted different movement strategies under dual-task conditions (McNevin and Wulf, 2002; Lacour et al., 2008). In the capacity sharing theory, when participants engage in dual-tasks, their attention shifts partly from the primary motor task to the supplementary task (in our study, a cognitive task). Consequently, cognitive resources are partially allocated to this additional task, the conscious control over body movements reduced (Lacour et al., 2008). To some extent, automatic and reflexive postural control strategies then take precedence to sustain motor performance (McNevin and Wulf, 2002; Lacour et al., 2008). This shift may result in superior motor performance compared to single-task conditions (Wulf et al., 2004; McNevin and Wulf, 2002; Vuillerme et al., 2000), avoiding the occurrence of injuries. For example, the lower ankle inversion angle and angular velocity observed in our study. A smaller ankle inversion angle indicates that the lateral ankle ligaments experience less strain due to inversion (Yıldız and Yalcın, 2013). Similarly, a lower ankle angular velocity indicates that muscles, such as the peroneal muscles, have adequate reaction time to activate and counteract excessive ankle inversion (Ashton-Miller et al., 1996), which all suggest a potential decrease in the risk of ankle sprains (Koshino et al., 2017).

In contrast to individuals without CAI, those with CAI did not show a decrease in injury potential when transitioning from single to dual-task conditions. The maladaptive neuroplastic changes at both spinal and cortical levels follows musculoskeletal injuries like CAI may be the reasons (Bruce et al., 2020), counteracting the injury potential that should be reduced under dual-task conditions. In the spinal level, the protective mechanism for protecting muscles from excessive strains after injuries, i.e., arthrogenic muscle inhibition (Norte et al., 2022), would affect the ability of voluntary muscle activation. This, in turn, may result in muscle weakness around ankle due to heightened activation of inhibitory interneurons synapsing in the motor neuron pool, thus reducing the efficiency of motoneuron recruitment by the CNS after injuries (Dong et al., 2024). Dong et al. demonstrated that arthrogenic muscle inhibition existed in peroneal muscles among individuals with CAI (Dong et al., 2024), which is the dominating muscles to prevent excessive ankle inversion (Ashton-Miller et al., 1996). The functions of the muscles may be inhibited by the protective mechanism at spinal level, leading to unchanged indicators concerning ankle inversion under dual-task conditions compared to single-task conditions among individuals with CAI. In the cortical level, numerous studies showed that the corticomotor excitability decreased in the projection areas of lower limb muscles around ankle in the M1 among individuals with CAI (Pietrosimone and Gribble, 2012; Terada et al., 2022; Nanbancha et al., 2019; Kosik et al., 2017), making them hard to activate the cortical motor neurons (Pietrosimone et al., 2012; McLeod et al., 2015), then leading to muscle weakness and failure of muscle activation (Pietrosimone et al., 2012). Activities of the secondary sensorimotor cortex and other cortical areas would increase to compensate for the declined cortical excitability in M1 to maintain motor performance (Needle et al., 2017; Bruce et al., 2020), but the cortical compensatory mechanism is vulnerable, which tends to collapse under dual-task conditions, leading to poorer motor performance and higher injury potential (Bruce et al., 2020). Similarly, the concurrent subtraction task may break the compensatory mechanisms at cortical level (Burcal et al., 2019), lead to the inability to confront ankle inversion from single-to dual-task conditions among individuals with CAI during landing.

There are three limitations to this study. First, participants were aware that the flippable platform surface of the trap-door device would flip during landing, so this was an anticipatory condition that may differ from a non-anticipatory condition when a real injury occurs. Second, the gender distribution was not uniform across the two groups in this study, which introduces a potential bias in the results due to gender imbalance. Considering that males and females utilize distinct feedforward control strategies during landing, with females typically activating their knee extensors earlier than males to mitigate the deficit in hip extensor rate tension development (Stearns-Reider and Powers, 2018; Di Giminiani et al., 2020), it is advisable for future research to maintain a balanced gender representation within subgroups to mitigate any potential confounding effects of gender on postural control outcomes. Last, this study did not include direct indicators of muscle activity and ligament status, and it is recommended that more attention be paid to neuromuscular factors in future studies to provide evidence through electromyographic measurements of muscle activity and computer simulation modeling calculations of ligament strain to better assess the potential of ankle injury.

5 Conclusion

Individuals with CAI have a reduced ability to limit ankle inversion, inferring increased susceptibility to ankle sprains. Under dual-task conditions, individuals without CAI limited their ankle inversion, while those with CAI did not, inferring a higher injury potential among those with CAI. Drop-landing, especially under dual-task conditions, poses a high risk of excessive ankle inversion for individuals with CAI.

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 humans were approved by the Institutional Review Board of Shandong Sport University (Approval Number: 2023014) and was in accordance with the Declaration of Helsinki. The studies were conducted in accordance with the local legislation and institutional requirements. The participants provided their written informed consent to participate in this study.

Author contributions

CZ: Methodology, Data curation, Software, Writing–original draft. XL: Writing–original draft, Formal Analysis, Investigation. HG: Writing–original draft, Methodology, Validation. TZ: Methodology, Validation, Writing–original draft. XZ: Writing–original draft, Data curation. XH: Formal analysis, Writing–original draft, Writing–review and editing. PS: Conceptualization, Funding acquisition, Methodology, Project administration, Writing–review and editing.

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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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Muscle power is associated with higher levels of walking capacity and self-reported gait performance and physical activity in individuals with cerebral palsy

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Introduction: The purpose of this study was to investigate the relationships between a Power Leg Press test (PLP) with walking capacity and self-reported performance and participation in individuals with Cerebral Palsy (CP), and to compare the strength of the associations between two power tests (PLP and isokinetic (IsoK)) with walking capacity.

Methods: Ambulatory individuals with CP (n = 33; age 17.89 $\pm$ 7.52 years) performed five inclined power leg presses at 40%–50% of their 1-repetition maximum "as fast as possible". A linear position transducer was attached to the weight bar, and the displacement, total load, and angle of the sled were used to calculate peak power for each trial. Isokinetic knee extensor power was measured at 60 deg/sec. Walking capacity was measured using the 10-m walk test fast (FS) and self-selected (SS) speeds and the 1-min walk test (1MWT). Self-reported performance and participation measures were the Activities Scale for Kids-performance (ASKp), Patient-Reported Outcomes Measurement Information System (PROMIS®), and the Gait Outcomes Assessment List (GOAL). Pearson's correlation coefficients determined relationships between power measures with walking capacity and self-report measures ($\alpha < 0.05$).

Results: PLP and IsoK power were significantly correlated to SS (r = 0.361, r = 0.376), FS (r = 0.511, r = 0.485), and 1MWT (r = 0.583. r = 0.443), respectively ($p < 0.05$). There was no significant difference between the strength of the associations between walking capacity and each test of power (PLP and Isok) ($p > 0.05$). PLP power was significantly correlated to composite scores on the ASKp (r = 0.690) and GOAL (r = 0.577) and to four components of the PROMIS, including physical function (r = 0.588) ($p < 0.01$). The Gait and Mobility subscale of the GOAL (r = 0.705) and the Locomotion (r = 0.636), Transfers (r = 0.547), and Standing (r = 0.521) subscales of the ASKp had strong relationships to peak power produced during the PLP test ($p < 0.01$).

Discussion: PLP power was significantly correlated with walking capacity and self-reported walking performance and mobility-based participation in ambulatory individuals with CP. Higher movement velocities associated with the PLP test may explain the significant associations of power with faster gait speeds. Self-reported mobility performance and physical activity also showed moderate to strong relationships with lower extremity power. Overall, these results suggest a strong link between decreased muscle power generation and walking limitations in individuals with CP.

KEYWORDS

muscle power, walking, activity, participation, gait speed, muscle performance

1 Introduction

Cerebral Palsy (CP) is a neurologic disorder resulting from injury to the brain prior to or shortly after birth (Rosenbaum et al., 2007) leading to significant and lasting impairments to the neuromuscular system including decreased muscle performance (weakness, decreased power and decreased rate of force development), spasticity and co-contraction (Damiano et al., 2001; Ross and Engsberg, 2007; Moreau et al., 2012; Moreau, 2013). Impairments in muscle performance have devastating impacts on individuals with CP as they influence walking and other critical mobility skills that are essential for maintaining independence and quality of life (Lundh et al., 2018). The World Health Organization's International Classification of Function and Disability (ICF, 2001) model is a biopsychosocial model that can be used to organize and describe the influence CP has across three domains and how these domains influence each other (2001). The three domains include: 1) body systems (parts of the body and their function), 2) activities (both *capacity* and *performance* of tasks by an individual), and 3) participation (function of an individual in all areas of life). For example, in those with CP, lower extremity muscle weakness (impairment in body system domain) limits walking ability (limitation in activity domain) which may result in difficulty participating in school and recreational tasks (participation restriction) (Pirpiris et al., 2003; Schenker et al., 2005; Morgan et al., 2014). Thus, understanding the muscle performance impairments which are most impactful on walking ability will be important in developing rehabilitation interventions that are functionally meaningful for those with CP.

Muscle strength and power are two elements of muscle performance that are critical to walking activity and participation in those with CP. Lower extremity muscle strength, or the ability to produce maximal force, is decreased in those with CP (Wiley and Damiano, 1998) and is associated with greater walking capacity in individuals with CP (Damiano et al., 2001). Muscle power is defined as the product of force and velocity and is more impaired than muscle strength in those with CP (Moreau, 2013; Moreau et al., 2013; Geertsen et al., 2015). In addition, high velocity (power) training has been shown to be effective at improving walking speed and functional walking performance in youth with CP (Moreau et al., 2013; Kara et al., 2023). Only one study has explored associations between power and walking capacity (Moreau and Lieber, 2022) and none have reported associations between power and self-reported activity and participation. Moreau et al. reported a moderate to good relationship between isokinetic knee extensor power with both fast gait speed ($r = 0.65$; $p < 0.001$) and the 1-min walk test (1MWT) ($r = 0.79$; $p < 0.001$) in a cohort of ambulatory individuals with CP (Moreau and Lieber, 2022), however this study used an isokinetic measure of power and evaluated these relationships retrospectively. To our knowledge, no studies have prospectively examined the associations between power and walking activity and participation in those with CP. Thus, additional studies are needed to understand how muscle power influences individuals with CP across the ICF continuum of activity (capacity and performance) and participation.

Muscle power is typically measured using isokinetic dynamometry, cycle ergometry, or with clinical field tests. While each of these measures have value, they present several clinical limitations. Isokinetic dynamometry, the gold-standard measure of muscle power, is costly and can be time consuming to set up and administer which can limit clinical feasibility. Similarly, cycle ergometry tests, like the Wingate Anaerobic Test (WAnT), require specialized computer equipment to measure lower extremity power which can also be cost prohibitive in most clinical settings (Bar-Or, 1987). Most recently, Verschuren et al. developed a field test to measure power, the Muscle Power Sprint test (MPST) (Verschuren et al., 2007). While this test is cost effective and clinically feasible, it requires individuals with CP to sprint which has limited use in those with lower mobility levels. Based on current evidence, there are few tests of lower extremity power that are clinically feasible, cost effective, and appropriate for individuals with CP with a wide range of mobility levels (Pontiff et al., 2021; Pontiff et al., 2023).

A power leg press (PLP) test was recently shown to be a valid and reliable measure of lower extremity power in typically developing individuals (Pontiff et al., 2021) as well as ambulatory adults and children with CP (Pontiff et al., 2023). The PLP test measures lower extremity muscle power during a closed-chain leg press activity on a standard piece of fitness equipment. Further, the PLP test also uses an adjustable inclined leg press and weight bar attachment to accommodate a wide range of mobility levels in those with CP. Thus, the PLP may be a useful test as it is clinically feasible, more cost-effective than isokinetic dynamometry, and adaptable to wide range of mobility levels in individuals with CP. However, associations to key functional tasks like walking capacity and participation have not been explored. Further, the strength of the associations to walking capacity in comparison to the gold standard measure of lower extremity power, isokinetic dynamometry has not been evaluated. Understanding the strength of these relationships will aid researchers and clinicians in selecting the measure of power with the closest association to function which can be used to measure change in patient performance after intervention.

Thus, the first aim of this study was to examine how lower extremity muscle power produced during a closed chain PLP test was associated with walking capacity and self-reported walking performance and participation in individuals with CP. We hypothesized that lower extremity muscle power will be positively correlated to tests of walking capacity and self-reported performance and participation. The second aim of this study was to compare the strength of the relationships between muscle power produced during the PLP test and a gold standard test of power, isokinetic dynamometry, with walking capacity. For the second aim, we hypothesized that the PLP test will demonstrate stronger relationships to walking capacity compared to isokinetic dynamometry because the PLP test is a closed chain movement requiring use of the hip and knee extensors as well as the plantarflexors, which are the same critical muscle groups used in gait. The isokinetic power test is an open chain task that isolates the knee extensors. Using clinically feasible power tests to explore key relationships between muscle power with walking capacity and self-reported activity and participation will guide rehabilitation providers and researchers in the development of important exercise interventions to target improvements in activity and participation for those with CP.

2 Materials and methods

Ambulatory adults and children with a diagnosis of CP (Gross Motor Function Classification System level I-III) between 10–40 years were recruited and eligible for participation in this prospective study. Participants were excluded if they met the following criteria: botulinum toxin injections in the lower extremities in the last 3 months, knee flexion contractures >25°, surgery on the lower extremities in the last year, or inability to ambulate. This study was approved by a university Internal Review Board (IRB). Study participants were recruited from physical therapy clinics, physician referral, community organizations, and from a randomized clinical trial (Clinicaltrials.gov: NCT 03625570). Informed consent and assent were obtained prior to testing from adult and child participants, respectively.

2.1 Testing procedures

Following consent, participants completed a brief health and surgical history in addition to a short physical exam which included: height, body mass, and leg length measures.

2.2 Power leg press (PLP) test

Specific testing procedures for the PLP test have been previously published (Pontiff et al., 2021; Pontiff et al., 2023), but a brief description is provided here. 1-Repetition Maximum (1-RM) testing was performed prior to the PLP test to determine the appropriate load for the power test. Methods for 1-RM testing for individuals with CP have been previously published (Pontiff and Moreau, 2022). Power testing loads were set at 40%–50% of the individual's 1-RM based on the testing guidelines for healthy adults (ACSM, 2019) and children (Faigenbaum et al., 2009). To begin, participants were positioned on the Total Gym® (Total Gym Fitness Studios, San Diego, CA) leg press with their knees flexed to 90° (measured with goniometer), feet flat on the foot plate and the hands on the handlebars with the elbows fully extended. This starting position was maintained for both the 1-RM and PLP test. Participants performed three to four practice trials where they pushed the load "as fast as possible" from 90 degrees of knee flexion to their full knee extension and returned slowly to the start position. Participants rested for 5 minutes and then performed five consecutive power leg presses. A linear position transducer (TE Connectivity®, SGD 120-in Cable Actuated Sensor; Chatsworth, CA) was attached to the weight bar. Displacement of the transducer cable during the concentric phase of the movement produced a voltage signal (sampling frequency 500 Hz) that was recorded and converted into position-time data by a custom LabVIEW® (National Instruments Corp., Austin, Tx) program. Power was calculated as the product of force and velocity using custom MATLAB® (Mathworks Inc., Natick, MA) code. Force (*F*) was defined as $F = m*g*sin(\theta)$, where m is the total mass of the system (body mass + external weight plates + mass of the sled in kg), g is acceleration due to gravity (9.81), and θ is equal to the angle of the leg press sled from the horizontal (29.4°) (Pontiff et al., 2021). Peak Power was calculated as the maximal power value in the concentric phase of the press. The mean of all valid trials from the test was normalized to participant body mass in kg and this value was used for all subsequent analyses. We selected mean peak power across trials because this variable was the most reliable and had the smallest minimal detectable change (MDC) in children and adults with CP compared to three other power measures in our validation study (Pontiff et al., 2023). Overall, measures of peak power had smaller MDC scores as compared to measures of average power for the PLP test (Pontiff et al., 2023). Repetitions were removed if the participant demonstrated a movement error or reduced effort as indicated by a value below 2SD of the mean of the five repetitions.

2.3 Isokinetic dynamometry

An isokinetic dynamometer (System 4, Biodex Medical Systems, Shirley, NY) measured knee extensor muscle power of the participants most involved limb at 60°/second. Participants were seated upright (trunk 85° from the horizontal) with their knee joint aligned with the axis of the dynamometer. The trunk, waist, and thighs were secured with straps, and the arms were crossed across the chest to minimize compensatory movements during testing. The range of motion of the knee was set through the full available passive range, and torque was gravity corrected. One to three practice trials were performed for warm up and familiarization. Following practice, each participant rested for 5 minutes.

For the isokinetic (IsoK) test, participants were asked to kick their leg into extension "as hard and as fast as possible" for five repetitions. Consistent verbal encouragement was given for each repetition. IsoK Power (IsoK P) was defined as the highest average power produced across the five repetitions during the concentric constant velocity portion of the knee extension movement at 60°/second, which was calculated by the computerized software of the isokinetic dynamometer (System 4, Biodex Medical Systems, Shirley, NY). Repetitions were removed if the participant demonstrated a movement error or reduced effort as indicated by a value below 2SD of the mean of the five repetitions at 60°/second. IsoK power data were normalized by body mass in kg for statistical analyses.

2.4 Measures of activity: walking capacity tests

Walking capacity was measured using the 10-m walk test (Fast and Self-selected: FS and SS speeds) and the 1-min walk test (1MWT). Participants wore comfortable clothes, walking shoes, braces or orthoses (if appropriate), and used any assistive devices they used for community walking.

For the 10-m walk test, participants were instructed to walk on a flat, level 14-m course in a hallway. The middle 10 m was timed for each test, so a 2-m acceleration and 2-m deceleration were marked to indicate where the timer was to start and stop, respectively. Participants were instructed to "walk at your normal comfortable walking speed until you reach the end of the hall" for the SS test. For the FS speed, individuals were instructed to "walk as fast as you safely can until you reach the end of the hall". Three trials were performed at both speeds, and the average time for three trials was divided by 10 m to calculate a FS and SS gait speed in meters per second (m/s). The 10-m walk test has been shown to be a valid and reliable measure of gait speed in those with CP (Graser et al., 2016; Bahrami et al., 2017).

Following the 10-m walk test, participants rested for 5 minutes and were given instructions for the 1MWT. Participants then walked as fast as possible without running for 1 min. Each participant was notified when there were 30 s, 10 s and 5 s remaining. When time expired, participants were instructed to "stop" and the floor was marked with tape of the farthest forward foot and the score was recorded as meters walked in 1 minute. The 1MWT has been shown to be a valid and reliable measure of walking capacity in individuals with CP (McDowell et al., 2009).

2.5 Measures of self-reported activity (performance) and participation: self-report questionnaires

The Performance version of the Activities Scale for Kids (ASKp), Patient-Reported Outcomes Measurement Information System (PROMIS®)-Pediatric Profile-49 v2.0, and the Gait Outcomes Assessment List (GOAL) were used to measure self-reported performance and participation. Parent and participant versions were used for the PROMIS and GOAL as appropriate. The ASKp is a 30-item self-reported measure with nine sub-domains (Young et al., 2000). Scores are expressed as a percentage of function with lower scores indicating less functional ability. The PROMIS Pediatric and Parent Proxy Profile instruments are a collection of six self-reported questionnaires with the following domains: Depressive Symptoms, Anxiety, Physical Function-Mobility, Pain Interference, Peer Relationships, along with a single item on Pain Intensity (DeWalt et al., 2015). The PROMIS is a valid and reliable measure for children with CP and the mobility domain is capable of distinguishing between Gross Motor Function Classification System (GMFCS) levels (Kratz et al., 2013). The GOAL is a 48-item self-reported questionnaire with seven domains that evaluate a child's performance related to gait function and mobility. The GOAL can discriminate between GMFCS levels and is a valid measure of gait function in ambulant children with CP (Thomason et al., 2018).

TABLE 1 Participant demographics.

Characteristics	All participants (n = 33)
Sex	
Male	12
Female	21
Age	
mean (SD), yrs	17.89 (7.52)
range, yrs	10–37
Height, mean (SD), m	1.52 (0.13)
Body Mass mean (SD), kg	51.22 (15.49)
Topographical Classification, n	
Hemiplegia	4
Diplegia	22
Triplegia	3
Quadriplegia	4
GMFCS Level, n	
I	8
II	22
III	3

Abbreviations: Gross Motor Function Classification System (GMFCS) level.

2.6 Statistical analysis

We used a pilot sample to perform an *a priori* power analysis. Based on a sample of 15 youth with CP, isokinetic knee extensor power was strongly associated with distance walked during the 1MWT ($r = 0.78$, $p < 0.05$). Assuming similar results for Aim 1, we determined that a minimum sample size of 30 achieves 90% power to detect a difference of 0.55 between the null hypothesis correlation of 0, and the alternative hypothesis correlation of 0.55 using a two-sided hypothesis test. For Aim 2, assuming that we will obtain a similar r value, a sample size of 30 would be enough to detect a difference of 0.16 between two tests' correlation coefficients with 82% power. Power increases to 100% when the difference is greater than 0.16.

The relationships between each power test (PLP and IsoK) with tests of walking capacity and self-reported activity and participation measures were examined using Pearson's correlation coefficients. Correlation coefficients were categorized based on previously published recommendations with values < 0.25 considered poor, 0.25–0.50 were weak, 0.5 to 0.75 were moderate to good, and coefficients >0.75 were excellent associations (Portney LG, 2000). Correlation coefficients were compared statistically to determine which test (PLP or IsoK) demonstrated a stronger relationship to walking capacity. To compare correlation coefficients, we used methods published by Steiger (1980). A computerized software program (Lee, 2013) converted correlation coefficients to z-scores using Fisher's transformation and the asymptotic covariance of the estimates were used for an asymptotic z-test to determine if the correlations were statistically different from one another. Significance levels were set at 0.05 for all analyses.

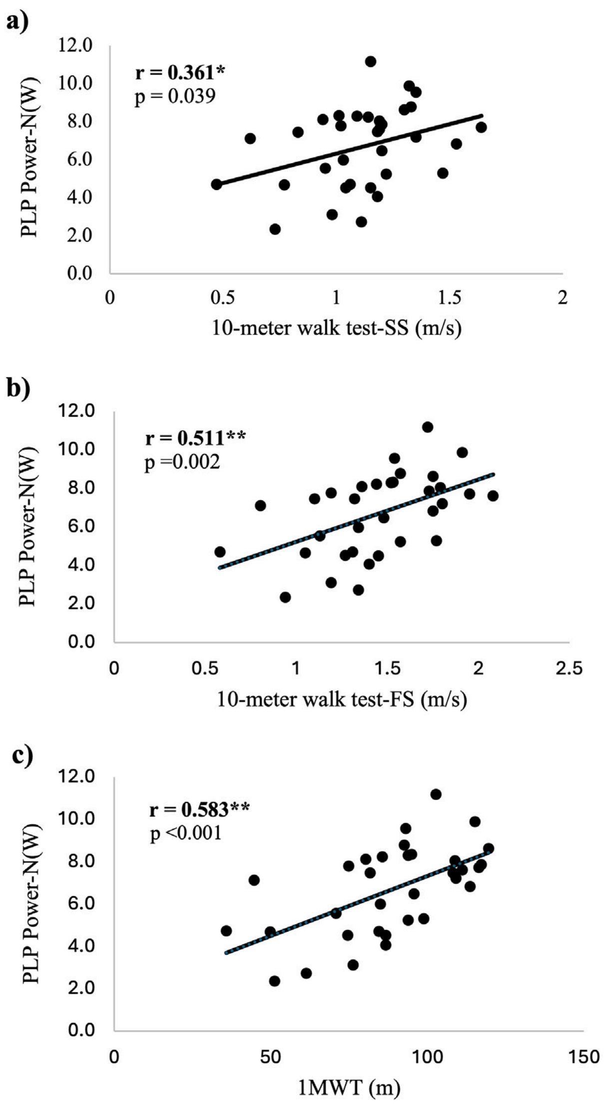

FIGURE 1
Scatter plots demonstrating the associations between normalized PLP power and tests of walking capacity. **(A)** association between PLP power and SS gait speed **(B)** association between PLP power and FS **(C)** association between PLP power and 1MWT. Abbreviations: PLP Power-N(W) = Power Leg Press Power in watts normalized by body mass; SS = Self-selected; FS = Fast speed; 1MWT = 1 minute walk test; *=statistical significance p<.05; **=statistical significance p<.01; sample size n = 33.

3 Results

Thirty-three individuals with CP participated in this study. Participant demographics are reported in Table 1.

3.1 Associations between lower extremity power and walking capacity

The average load for the PLP test for all participants was 173.6 lbs ±95.5 lbs with a range of 15lbs–410 lbs. This refers to the amount of external weights added to the weight bar, unadjusted for the angle of the sled. The average %1RM across all participants for the PLP test was 47.3% ± 6.8%. PLP peak power was significantly and positively associated with walking speed (SS and FS) and 1MWT ($p < 0.05$) with the strength of the associations ranging from weak to moderate (Figure 1).

3.2 Comparison of correlations for associations between PLP and isokinetic tests for measures of walking capacity

There were no significant differences between correlation coefficients for the PLP and IsoK power tests and any of the walking tests ($p > 0.05$ – see Table 2). In addition, scatter plots depicting the associations between PLP power and walking capacity, and IsoK power and walking capacity are displayed in Figures 1, 2, respectively.

3.3 Associations between lower extremity power with self-reported activity (performance) and participation

PLP Power was significantly associated with the composite score as well as each subcomponent of the ASKp ($p < 0.05$). All components of the ASKp were moderately correlated with PLP power (r = 0.484–0.690, $p < 0.05$). The PROMIS domains of physical function and mobility, fatigue and peer relationships were moderately correlated PLP power (r = −0.524 to 0.588, $p < 0.05$), while pain intensity demonstrated a weak correlation with PLP power (−0.257, $p < 0.05$). All components of the GOAL were significantly correlated with PLP power (r = 0.361–0.705, $p < 0.05$) except Body Image and Pain ($p > 0.05$). The strength of the correlations for the components of the GOAL ranged from weak to strong. Correlation coefficients between PLP power and all self-reported questionnaires can be found in Table 3.

4 Discussion

This study aimed to 1) explore the associations between lower extremity muscle power during a PLP test with walking capacity and self-reported walking performance and participation in individuals with CP, and 2) to compare the strength of the associations between two different tests of muscle power and walking capacity. Lower extremity muscle power was significantly related to walking capacity and self-reported walking performance and participation in ambulatory individuals with CP which supports our first hypothesis. Both PLP and IsoK power were significantly related to all three tests of walking capacity. There was no statistically significant difference in the associations of muscle power with walking capacity between the two power tests which refutes our second hypothesis. However, the strength of the associations between power and FS speed and the 1MWT were moderate for PLP power and weak for IsoK power.

TABLE 2 Comparison of correlation coefficients for associations between PLP and IsoK Power tests and walking capacity.

Association Tested $r_{PLP\text{-}walking\ test}/r_{IsoK\text{-}walking\ test}$	Correlation coefficients
$r_{PLP\text{-}SS\ speed}/r_{IsoK\text{-}SS\ speed}$	0.361/.376 $p = 0.904$
$r_{PLP\text{-}FS\ speed}/r_{IsoK\text{-}FS\ speed}$	0.511/.485 $p = 0.822$
$r_{PLP\text{-}1MWT}/r_{IsoK\text{-}1MWT}$	0.583/.443 $p = 0.214$

Abbreviations: r = Pearson's r correlation coefficient; PLP- power leg press power; IsoK- isokinetic power; SS- self selected; FS- fast; 1MWT- 1-min Walk Test; sample size n = 33.

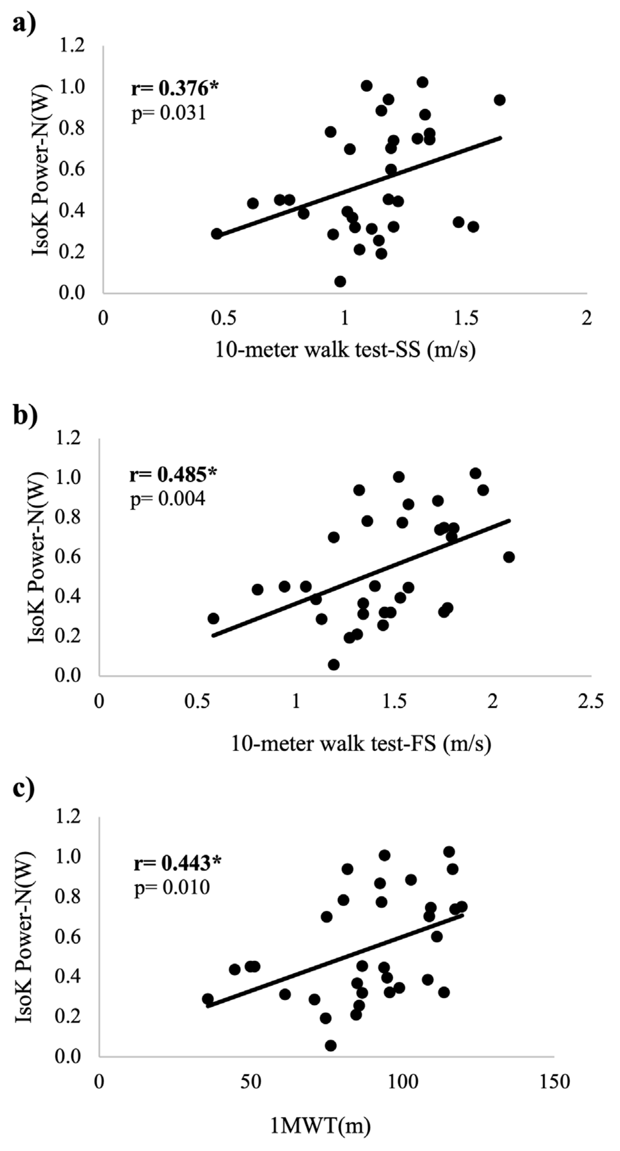

FIGURE 2
Scatter plots demonstrating the associations between normalized IsoK power and tests of walking capacity. **(A)** association between IsoK power and SS gait speed **(B)** association between IsoK power and FS **(C)** association between IsoK power and 1MWT. Abbreviations: IsoK Power- N(W) = Isokinetic Power in Watts normalized by body mass; SS = Self-selected; FS = Fast speed; 1MWT = 1 minute walk test; *=statistical significance p<.05; **=statistical significance p<.01; sample size n = 33.

4.1 Associations between lower extremity power and walking capacity

Walking capacity was significantly correlated with peak power produced during a PLP test in ambulatory individuals with CP, which is consistent with our primary hypothesis. Both the 1MWT and FS speed had stronger associations with peak power (moderate associations: r = 0.583, r = 0.511; $p < 0.01$) compared to SS speed (weak association: r = 0.361, $p < 0.05$) (Figure 1). One consideration for the stronger relationship with 1MWT and FS speed compared to SS speed is the participant's effort to walk "as fast as possible" for these fast walk tests. This effort is similar to the PLP test which asked individuals to press "as fast as possible" through a squat-like motion in order to promote power generation. Conversely the SS speed asks participants to walk at their usual pace. The similarities in effort of performance and velocity of movement between the PLP and both fast walking tests (FS speed and 1MWT) may explain the stronger associations we reported with these tests as compared to the correlation between SS speed and peak power. The 1MWT evaluates the farthest distance walked in 1 minute at the fastest speed possible. Compared to the other three walking tests in this study, the 1MWT best reflects the components of power (force and velocity) and thus, demonstrated the strongest relationship to power (r = 0.583, $p < 0.01$).

4.2 Comparison of correlations for associations between PLP and isokinetic tests for measures of walking capacity

The goal of this secondary analysis was to compare the strength in associations between a newer test of power, the PLP test, and a gold standard measure of power, the IsoK test. Both power tests (PLP and IsoK) were significantly and positively correlated to all measures of walking capacity (SS and FS speed; 1MWT) (Figures 1, 2, respectively). When we compared the strength of the associations between the two tests, there were no significant differences (Table 2). One retrospective study explored the associations between isokinetic average power and walking capacity in individuals with CP (Moreau and Lieber, 2022), and the strength of these relationships was stronger than our data for both tests of power. A moderate to good relationship between isokinetic knee extensor power at 60°/second and both FS gait speed (r = 0.65; $p < 0.001$) and the 1MWT (r = 0.79; $p < 0.001$) was reported in this larger cohort of ambulatory individuals with CP (Moreau and Lieber, 2022). In addition, the authors reported a moderate relationship between SS gait speed and isokinetic knee extensor power (r = 0.59; $p < 0.01$). There are no previously published studies exploring the associations between leg press power and walking capacity in individuals with CP for comparison. The differences in the strength of the correlations between the studies may, in part, be explained by the larger sample (n = 66) and narrower age range (5–25 years) (Moreau and Lieber, 2022) than our study (n = 33; age range = 10–37 years), and different methodologies, particularly for the PLP test. However, we recommend the PLP test for clinical testing of lower extremity power in individuals with CP because it involves multiple muscle

TABLE 3 Correlation coefficients between self-reported activity (performance) and participation and PLP Power in individuals with CP.

Self-report outcome measure	Correlation coefficient with PLP power
ASKp-Composite	0.690**
ASKp-Physical Component	0.484**
ASKp-Locomotion	0.636**
ASKp-Standing Skills	0.521**
ASKp-Transfers	0.547**
ASKp-Play	0.480**
ASKp-Dressing	0.590**
ASKp-Other	0.463**
PROMIS-Physical Function Mobility	0.588**
PROMIS-Fatigue	-.524**
PROMIS-Anxiety	-.268
PROMIS-Depression	-.180
PROMIS-Peer Relationships	0.457**
PROMIS-Pain Interference	-.333
PROMIS-Pain Intensity	-.257*
GOAL-Composite	0.577**
GOAL-ADL	0.478**
GOAL-Gait and Mobility	0.705**
GOAL-Physical Activity and Sport	0.495**
GOAL- Pain	0.243
GOAL- Gait Pattern and Appearance	0.361*
GOAL- Body Image	0.147

Abbreviations: PLP- power leg press; ASKp-Activities Scale for Kids-performance version; PROMIS- Patient-Reported Outcomes Measurement Information System; GOAL- gait outcomes assessment list; ADL-activities of daily living; * = statistical significance $p < .05$; ** = statistical significance $p < .01$; sample size n = 33.

groups in a closed chain fashion, is more cost-effective compared to an IsoK testing device, does not constrain the velocity as in isokinetic testing, and is conducted with equipment that is familiar to most practicing clinicians (Pontiff et al., 2023).

4.3 Associations between lower extremity power with self-reported activity (performance) and participation

Lower extremity peak power was moderately related to most self-reported measures of mobility-based performance and participation in ambulatory individuals with CP. All components of the ASKp and most components of the GOAL, including the composite scores, demonstrated moderate to strong relationships with lower extremity power. Of clinical significance was the strong relationship between power and the GOAL-Gait and Mobility (r = 0.705, $p < 0.01$, Table 3) and the moderate association with ASKp-Locomotion (r = 0.636; $p < 0.01$, Table 3) and the PROMIS Physical Function Mobility (r = 0.588; $p < 0.01$, Table 3). Moderate associations were also reported between PLP peak power and the ASKp- Transfers, standing skills, and dressing subscales (r = 0.521–0.590; $p < 0.01$, Table 3). The strength of these associations may be explained by the rapid movements required to walk and complete other mobility activities like stair climbing, transfers, and walking. Knee extensor angular velocities range from 186–230°/second for stair climbing and sit to stand tasks (Schenkman et al., 1996; Hortobágyi et al., 2003), and 257–357°/second during terminal swing of walking (Mentiplay et al., 2018), emphasizing the need for velocity with these types of mobility activities. In addition, Van Vulpen et al. demonstrated significant improvements in self-reported mobility on both the Goal Attainment Scaling (GAS) measure as well as the parent-reported mobility questionnaire (MobQues) following functional power training in children with CP (van Vulpen et al., 2018). Therefore, lower extremity power may be a key element in performance of gait and other critical mobility activities in those with CP.

All domains of the PROMIS were significantly correlated with peak power except for scales related to pain interference, anxiety, and depression (Table 3). Anxiety and depression are more prevalent in those with CP than TD peers and both have shown some associations to physical activity (Whitney et al., 2019). However, a variety of physical and non-physical risk factors have been identified as contributors to anxiety and depression in those with CP including: sleep, developmental comorbidities, communication problems, pain, problems with social development, fatigue, general physical activity and mobility restrictions (Whitney et al., 2019). Therefore, the multifactorial nature of these psychological constructs may explain the lack of association with peak power and these domains of the PROMIS. Pain interference is a complex construct that is influenced by physical, structural, cognitive, and psychological elements in those with CP (McKinnon et al., 2019). While pain has been documented to frequently interfere with self-care, sleep, quality of life, and attention (Ramstad et al., 2011; Ostojic et al., 2020), little is known about how pain interference influences specific muscle performance measures like strength and power (McKinnon et al., 2019). Studies reporting frequent pain interference with sleep, self-care, and quality of life in individuals with CP reported moderate pain intensity levels (Ramstad et al., 2011; Ostojic et al., 2020). When we examine our study sample, the average pain intensity for our cohort was 2.7 ± 3.1 with 14 of 33 individuals reporting 0 pain in the past 7 days and the average pain interference score was 48.1 ± 12.2. The low pain intensity and lower than average pain interference scores may partly explain why pain interference was not significantly associated with peak power.

The peer relations, fatigue, and pain intensity domains of the PROMIS were also significantly related to PLP peak power (Table 3). Lower levels of fatigue and pain have been associated with higher levels of walking ability and function in those with CP (Opheim et al., 2009). There was also a moderate association between power and peer relationships. Despite limited published work, impaired walking ability has been linked to reduced social support and peer relationships in children and adolescents with CP (Colver et al., 2015). This evidence suggests there is some relationship between physical performance and relationships with peers in those with CP. Thus, individuals with CP who have greater deficits in muscle performance may perceive lower levels of peer acceptance and quality of friendships.

4.4 Limitations

There are several limitations that should be acknowledged in this study. First, the self-reported questionnaires used in this study are designed and validated in children with CP. While children were the majority of the sample in this study, 10 adults with CP participated in the study. Despite this limitation, we still believe that the information collected was valid as the questionnaires focused on mobility tasks that are relevant for both adults and children with CP. For example, when we examine the Gait and Mobility sub-section of the GOAL, which was strongly correlated to power, participants were asked about the difficulty with getting around their home, walking for more than 15 min, going up and down stairs, and walking faster than usual to keep up with others. These are critical daily tasks for adults with CP, and the associations with power may give us important information about how this element of muscle performance impacts activities of daily living in adults and youth with CP. Secondly, our sample included individuals with CP with different sexes and a wide age range. While a heterogenous sample can limit conclusions about more homogenous subgroups of people with CP, we sought to recruit a diverse sample so that the findings can be more generalizable to a larger group of individuals with CP. Third, impairments in posture, dynamic balance, and motor control are known to influence the ability to walk in individuals with CP (Rethwilm et al., 2021). The focus of this study was on the influence of muscle power on walking capacity; however, the impacts of balance and motor control should be evaluated in future investigations. Lastly, this study evaluated correlations between power, walking capacity and self-reported performance and participation. Correlation studies can describe associations or relationships between variables but cannot determine cause and effect. Despite the inherent limitations of correlational studies, these data may help to inform more robust randomized clinical trials aimed at determining the effectiveness of an intervention targeting lower extremity muscle power in improving walking activity and participation outcome measures.

5 Conclusion

This study examined the associations between lower extremity power produced during a closed chain leg press task with walking activity and participation in ambulatory individuals with CP. Our findings demonstrate that leg press power is significantly and positively associated with walking capacity and self-reported walking performance and mobility-based participation in ambulatory individuals with CP. For measures of walking capacity, the distance walked in 1 minute (1MWT) and FS walking speed had the strongest associations with PLP power. These fast walking tests may be more strongly linked to peak power generation as they utilize both components of power–force and velocity. Further, muscle power was significantly related to walking capacity in individuals with CP regardless of which method was used to measure muscle power. However, we recommend the PLP test as it incorporates multiple lower extremity muscle groups and is more functional and cost-effective compared to isokinetic dynamometry. Self-reported mobility performance and physical activity outcomes also showed moderate to strong relationships with lower extremity muscle power in subscales involving activities that require more rapid movements, such as gait, transfers, and physical activity. Overall, these findings highlight the critical associations between lower extremity power with walking capacity and mobility-based performance and participation, providing evidence of a strong link between decreased muscle power generation and walking limitations in individuals with CP. Based on the findings of this study, we recommend that clinicians include measures of lower extremity power in their clinical evaluations as power is related to meaningful measures of activity and participation in those with CP. The significant associations between power with walking capacity and self-reported participation highlight activity limitations and participation restrictions that may improve with interventions targeting muscle power generation in ambulatory individuals with CP. The effectiveness of power training interventions on walking capacity, performance, and mobility-based participation should be evaluated in future work.

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 humans were approved by Louisiana State University Health Sciences Center- New Orleans Internal Review Board. The studies were conducted in accordance with the local legislation and institutional requirements. Written informed consent for participation in this study was provided by the participant's legal guardians/next of kin.

Author contributions

MP: Conceptualization, Data curation, Formal Analysis, Funding acquisition, Investigation, Methodology, Project administration, Writing–original draft. AB: Data curation, Formal Analysis, Methodology, Writing–review and editing. LL: Conceptualization, Methodology, Supervision, Writing–review and editing. NM: Conceptualization, Data curation, Formal Analysis, Funding acquisition, Investigation, Methodology, Project administration, Supervision, Writing–review and editing.

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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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Optimal vocal therapy for respiratory muscle activation in patients with COPD: effects of loudness, pitch, and vowels

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Background: Vocal therapy, such as singing training, is an increasingly popular pulmonary rehabilitation program that has improved respiratory muscle status in patients with chronic obstructive pulmonary disease (COPD). However, variations in singing treatment protocols have led to inconsistent clinical outcomes.

Objective: This study aims to explore the content of vocalization training for patients with COPD by observing differences in respiratory muscle activation across different vocalization tasks.

Methods: All participants underwent measurement of surface electromyography (sEMG) activity from the sternocleidomastoid (SCM), parasternal intercostal muscle (PARA), seventh intercostal muscle (7thIC), and rectus abdominis (RA) during the production of the vowels/a/,/i/, and/u/at varying pitches (comfortable, +6 semitones) and loudness (–10 dB, +10 dB) levels. The Visual Analog Scale (VAS) was used to evaluate the condition of patients concerning vocalization, while the Borg-CR10 breathlessness scale was utilized to gauge the level of dyspnea following the task. Repeated-measure (RM) ANOVA was utilized to analyze the EMG data of respiratory muscles and the Borg scale across different tasks.

Results: Forty-one patients completed the experiment. Neural respiratory drive (NRD) in the SCM muscle did not significantly increase at high loudness levels (VAS 7-8) compared with that at low loudness levels (F (2, 120) = 1.548, P = 0.276). However, NRD in the PARA muscle (F (2, 120) = 55.27, P< 0.001), the 7thIC muscle (F (2, 120) = 59.08, P < 0.001), and the RA muscle (F (2, 120) = 39.56, P < 0.001) were significantly higher at high loudness compared with that at low loudness (VAS 2-3). Intercostal and abdominal muscle activation states were negatively correlated with maximal expiratory pressure (r = –0.671, P < 0.001) and inspiratory pressure (r = –0.571, P < 0.001) in the same loudness.

Conclusion: In contrast to pitch or vowel, vocal loudness emerges as a critical factor for vocalization training in patients with COPD. Higher pitch and loudness produced more dyspnea than lower pitch and loudness. In addition, maximal expiratory/inspiratory pressure was negatively correlated with respiratory muscle NRD in the same loudness vocalization task.

KEYWORDS

chronic obstructive pulmonary disease, vocalization training, neural respiratory drive, respiratory muscle, surface electromyographic

1 Introduction

Chronic obstructive pulmonary disease (COPD) is a heterogeneous lung condition characterized by chronic respiratory symptoms (dyspnea, cough, sputum production, and/or exacerbations) due to abnormalities of the airways and/or alveoli that cause persistent airflow obstruction (Celli et al., 2022). According to the estimation of large-scale epidemiology research, the global epidemic rate of COPD is 10.3% [95% confidence interval (CI) = 8.2%–12.8%], with the progress of the global population aging, the prevalence of chronic obstructive pulmonary disease will continue to increase (Adeloye et al., 2022; Adeloye et al., 2015). Although pulmonary damage in COPD is permanent, symptoms, such as respiratory muscle weakness and dyspnea, can be improved through pulmonary rehabilitation (Nolan et al., 2022).

Vocalization, such as singing, as a pulmonary rehabilitation program can combine specific abdominal respiratory patterns with respiratory muscle training to provide positive expiratory pressure and improve lung dynamic suction (O'Donnell and Webb, 2008; Kaasgaard et al., 2022; McNamara et al., 2017). Vocalization training can enhance exhalation muscle strength and FEV_1 in patients with COPD(McNamara et al., 2017; Bonilha et al., 2009). However, variations in vocalization treatment protocols have led to inconsistent clinical outcomes (Lord et al., 2012). Developing and applying effective vocalization mechanisms face challenges, including lack of consistent research content and a standardized vocalization protocol (Fang et al., 2022).

In addition, studies have shown significant differences in subglottal pressure during the vocalization of different vowels, suggesting that vowels may produce different afterloads in respiratory muscles (Pettersen, 2005). On the other hand, the activation of respiratory muscles varies with different loudness and pitch vocalization levels (Wang and Yiu, 2023). Hence, pitch, loudness, and vowels may be the main factors that contribute to differences in respiratory muscle activation during vocal content (Higgins, Netsell, and Schulte, 1998). In addition, unlike relaxed, natural expiration, vocalization requires the coordination of expiratory muscles and tends to cause dyspnea (Pettersen and Westgaard, 2004). Respiratory muscles, such as abdominal muscles, are the driving force for the power of vocalization. Although it is usually considered an inspiratory muscle, SCM is activated during speech (Wang and Yiu, 2023). Neural respiratory drive (NRD) measured by surface electromyography (sEMG) is a noninvasive measure of respiratory muscle activation that can be used in studies of physiologic mechanisms of respiratory muscles (Lin et al., 2019; Suh et al., 2020; AbuNurah, Russell, and Lowman, 2020).

Understanding the NRD of respiratory muscles in different vocalization tasks helps develop vocalization training programs for patients with COPD (Jolley et al., 2015; Pozzi et al., 2022). We aimed to 1) monitor the NRD of the SCM, PARA, 7thIC, and RA during various vocalization tasks by using sEMG; 2) evaluate the dyspnea index across different vocalization tasks; and 3) identify factors associated with respiratory muscle activation. We hypothesized that target muscles are more activated at high pitch and loudness and show different activity levels in vowel control tasks.

2 Materials and methods

From April 2023 to June 2023, we conducted a non-blind observational study in Qingdao, China. This study was registered at ChinaTrials.gov under the identifier ChiCTR2100052874 and was approved by the Ethics Committee of Qingdao Municipal Hospital. We posted the COPD pulmonary rehabilitation poster in several medical facilities to recruit participants. We screened other patients referred to outpatient clinics for pulmonary rehabilitation to determine their eligibility. The observation experiment occurred after formal recruitment in the intervention study, during their pulmonary rehabilitation sessions. Before participation, all patients were provided informed consent by signing a written document confirming their complete understanding of the study's purpose, procedures, and potential risks or benefits. The inclusion criteria for study patients encompassed a diagnosis of COPD based on the Global Initiative for Chronic Obstructive Lung Disease (GOLD), willingness to participate in the group, and normal vocal function. Patients with unstable heart diseases or severe cognitive impairment were excluded from the study.

2.1 Experimental protocol

After instructing patients in vocal techniques (pitch and loudness), the vowels of /a/,/i/, and/u/were assessed at varying pitch and loudness levels through a visual analog scale (VAS). Task1: To observe different vocal pitches, patients were instructed to relax their whole body and pronounce /a/,/i/, and/u/at low (VAS 2-3) and high pitches (VAS 7-8) with the same loudness (Bane et al., 2023; Awan et al., 2013; Awan et al., 2012). Task 2: To observe different vocal loudness, patients were instructed to pronounce /a/,/i/, and/u/at low (VAS 2-3) and high loudness (VAS 7-8) with the same pitches. Task 3: The same vocalization loudness of 60 dB was selected, and the patient was instructed to continue for more than 5 s to reach the tension-time threshold. The visual feedback interface displays real-time loudness. The patients had rest time between tasks and communicate fully to one another to ensure that patients are relaxed before pronouncing. The Borg-CR10 breathlessness scale was used to assess the degree of dyspnea of the patients after each

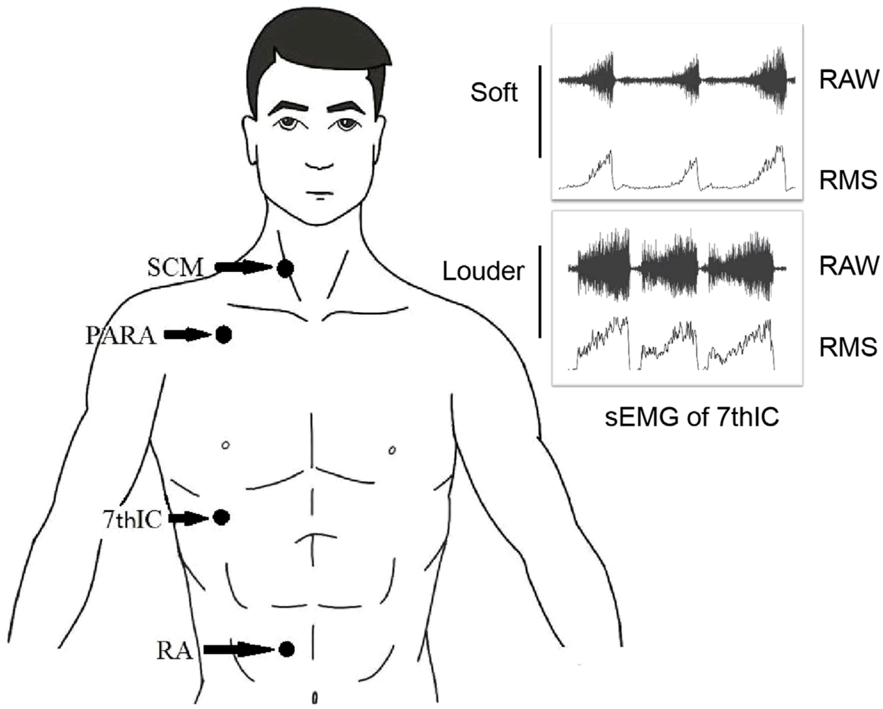

FIGURE 1
Electrode position and sEMG signals processing. Abbreviations: SCM, sternocleidomastoid; PARA, parasternal intercostal muscle; 7thIC, seventh intercostal muscle; RA, rectus abdominis; RAW, raw surface electromyographic signal; RMS, root-mean-square of surface electromyographic signal.

cycle of /a/,/i/,/u/. Two tasks were carried out at different periods, and the patients were assured of adequate rest before each task. A decibel meter measured at least 15 dB between low and high levels for accurate loudness difference. To assess different pitches, the patients initially produced a comfortable pitch while maintaining a comfortable loudness and then shifted to a higher base pitch (at least +5 semitones) while monitoring the pitch by using online tuning software (Wang and Yiu, 2023). Throughout the process, the voice was required to maintain a stable loudness and pitches and measured approximately 20 cm away from the participant. Data acquisition concluded when the voice reached a weak level (decrease of more than 5 dB or 2 semitones).

2.2 Surface EMG protocol

The Delsys Trigno™ wireless system (Delsys, Natick, MA) and four attached electrodes captured EMG signals at a sampling rate of 2,000 Hz. Wireless electrodes were placed at specific anatomical locations to ensure accurate measurements: the middle and lower 1/3 of the SCM, the junction of the PARA, the 7thIC, and 3 cm above the umbilicus in the position of the RA (Figure 1). Before electrode placement, the skin was meticulously cleaned with a medical alcohol pad to ensure optimal signal acquisition (Liu et al., 2019; da Fonsêca et al., 2019; Cabral et al., 2018). All surface EMG electrodes were positioned on the right side of the body for consistency (Ramsook et al., 2016). After data acquisition, all sEMG signals were analyzed using EMG works analysis software (Delsys, Natick). The sEMG signals recorded were filtered by a 20–450 Hz band-pass Butterworth filter. The signals were then segmented using a root-mean-square (RMS) value calculated based on a 100 ms moving window (Stepp, 2012) (Figure 1). The sEMG signal was calibrated as a percentage of the sEMG signal at maximum voluntary contraction (MVC) (Nguyen et al., 2020). NRD was used to represent RMS%MVC. During the performance of the MVC task, the subjects were instructed to exert their best effort by conducting three maximum breath tests (Cabral et al., 2018).

2.3 Statistical analysis

Statistical analysis used SPSS software v26.0 and GraphPad Prism v9.5.1. Data with normal distribution were presented as mean ± standard deviation (SD), while data with a skewed distribution were expressed as median ± interquartile range (IQR). Repeated-measures analysis of variance (RM-ANOVA) was utilized to analyze the EMG data of respiratory muscles and Borg scale across different tasks. Correlation analysis was conducted using Pearson correlation. A 95% confidence interval was established, and the significance level was set at 0.05.

3 Results

Forty-eight patients who met the criteria were recruited and trained; five were excluded because they could not maintain pitch or loudness, and two were excluded because they could not support sound. Finally, forty-one patients successfully concluded

TABLE 1 Basic information of subjects.

Subjects	41
Age (year)	66.3 (6.0)
Sex	
Male (%)	32 (78.0%)
Female (%)	9 (22.0%)
Height (m)	1.7 (0.1)
Weight (kg)	75.2 (12.5)
BMI (kg·m^{-2})	26.0 (3.7)
Pack years	40 (10)
FEV_1%pred	50.3 (15.0)
GOLD classification	
Class 1 (%)	8 (19.5%)
Class 2 (%)	22 (53.6%)
Class 3 (%)	11 (26.8%)
Class 4 (%)	0
MEP (mmHg)	68.7 (17.7)
MIP (mmHg)	81.0 (21.6)

Note: Data are presented as median (IQR) or n (%) unless otherwise stated; BMI, body mass index; FEV_1% pred, forced expiratory volume in one second % predicted; GOLD, global initiative for chronic obstructive lung disease criteria; MEP, maximal expiratory pressure; MIP, maximal inspiratory pressure.

the experiment. No adverse events occurred during the experiment. Demographic and anthropometric data of patents with COPD patients who underwent voice tasks are shown in Table 1.

3.1 NRD of muscles in pitch control

We found no difference at the NRD of SCM (F (2, 120) = 0.116, P = 0.890), PARA (F (2, 120) = 0.034, P = 0.967), 7thIC (F (2, 120) = 0.755, P = 0.473), and RA (F (2, 120) = 0.019, P = 0.982) during the high pitches task compared with the low pitch task (Table 2).

3.2 NRD of muscles in loudness control

The study showed no significantly higher NRD in patients with high loudness compared to those with low loudness in SCM (F (2, 120) = 1.300, P = 0.276) (Figure 2A). Furthermore, COPD patients with high loudness exhibited significantly higher NRD in the PARA (F (2, 120) = 55.27, P < 0.001) (Figure 2B), 7thIC (F (2, 120) = 59.08, P < 0.001) (Figure 2C), and RA (F (2, 120) = 39.56, P < 0.001) (Figure 2D) when compared to those with low loudness.

3.3 NRD of muscles in vowel control

No differences in NRD were found for SCM (F (2, 120) = 0.309, P = 0.735), PARA (F (2, 120) = 0.058, P = 0.944), 7thIC (F (2, 120) = 0.051, P = 0.944), and RA (F (2, 120) = 2.128, P = 0.124) across vowel tasks (Table 2).

3.4 Borg breathlessness score in different tasks

Borg dyspnea scores significantly differed across tasks [F (3, 160) = 66.47, P< 0.001]. We found no statistically significant difference in Borg dyspnea scores under low pitch and low loudness. The difference between Borg dyspnea scores under the high sound task was not statistically significant. However, Borg dyspnea scores were significantly higher after the high-loudness task than after the low-loudness task (P< 0.001) and after the low-pitched task (P< 0.001). Borg dyspnea scores were also significantly higher after the high-pitched task than after the low-loudness task (P< 0.001) and after the low-pitched task (P< 0.001) (Figure 3).

3.5 Correlations

Correlation analysis of expiratory muscle RMS measured after harmonization of pitch with baseline patient data showed that loudness was unrelated to baseline patient status. However, the RMS of the loudness control task was significantly negatively correlated with MEP (r = −0.671, P < 0.001) and MIP (r = −0.571, P < 0.001). Higher respiratory muscle strength was associated with lower RMS (Figure 4).

4 Discussion

Vocalization, such as singing training, can enhance expiratory muscle strength and improve lung function in patients with COPD (McNamara et al., 2017; Bonilha et al., 2009). However, its clinical effectiveness remains inconsistent and warrants further exploration due to the limited research on prescribing vocalization as a treatment (Fang et al., 2022). The neural drive of the expiratory muscles was significantly higher during high-loudness sounds compared with that during low-loudness sounds, and no differences were observed across varying pitches and vowel states. Second, high pitch and loudness produced higher dyspnea compared with low loudness and pitch. In addition, patients with higher MEP/MIP had lower respiratory muscle activation during the fixed loudness task.

4.1 NRD of respiratory muscle under different pitches and vowels

Categorization of vowels into high vowels and low vowels, along with mid vowels, is a well-established concept in linguistics. Subglottic pressure is produced by respiratory and laryngeal muscles, which is necessary for voice change (Traser et al., 2020). However, no significant difference was found in the NRD of

TABLE 2 Comparison of muscle activation in different task states.

Muscle/Task	Detailed characterisation of Tasks			*F*	*P*
Pitch control	Low	Medium	High		
SCM	0.46 (0.14)	0.47 (0.10)	0.47 (0.14)	0.116	0.890
PARA	0.48 (0.11)	0.47 (0.09)	0.47 (0.14)	0.034	0.967
7thIC	0.54 (0.10)	0.53 (0.07)	0.51 (0.09)	0.755	0.473
RA	0.51 (0.08)	0.51 (0.07)	0.51 (0.10)	0.019	0.982
Loud control	Soft	Medium	Louder		
SCM	0.43 (0.13)	0.47 (0.10)	0.46 (0.14)	1.300	0.276
PARA	0.37 (0.10)	0.59 (0.11)	0.47 (0.09)	55.27	**<0.001**[a]
7thIC	0.41 (0.10)	0.53 (0.07)	0.63 (0.10)	59.08	**<0.001**[a]
RA	0.41 (0.09)	0.50 (0.08)	0.58 (0.10)	39.56	**<0.001**[a]
Vowel control	/a/	/i/	/u/		
SCM	0.46 (0.09)	0.46 (0.09)	0.47 (0.10)	0.309	0.735
PARA	0.48 (0.08)	0.47 (0.08)	0.47 (0.07)	0.058	0.944
7thIC	0.52 (0.08)	0.52 (0.07)	0.52 (0.07)	0.051	0.950
RA	0.52 (0.08)	0.50 (0.07)	0.49 (0.07)	2.128	0.124

Note: SCM, sternocleidomastoid; PARA, parasternal intercostal muscle; 7thIC, seventh intercostal muscle; RA, rectus abdominis.
[a]$P< 0.05$ indicates a statistical difference.

respiratory muscle under different pitches and vowels. Plexico and Sandage (2012) showed that under any evaluation frequency, the threshold pressure value was not significantly different among the three consonant-vowel sequences, similar to the present results. However, Pettersen et al. found significant differences in the activation of intercostal, lateral abdominal, and rectus abdominus with different pitch and loudness levels. The vocalization task in Pettersen's study did not control for confounding factors, such as sound loudness. By contrast, loudness was controlled in this study, which may be the reason for the difference in the results (Pettersen, 2005). In addition, high-pitched voices produced prokinetic symptoms despite no significant respiratory muscle activation. Correspondingly, Wang and Yiu (2023). Showed that the activation level of suprhyal muscle was different with different vowels due to the different shapes of the mouth and the position of the tongue. Recruitment of laryngeal muscles increased significantly with increased pitch (Zhu et al., 2022). This finding suggests that perilaryngeal muscles rather than respiratory muscles produced pitch changes. This phenomenon is consistent with the view that the original power of the respiratory muscle produces sound and that the larynx and mouth are mainly used to modify the sound (Körner and Strack, 2023; Herbst, 2017). In conclusion, this study confirms that different pitches and vowels in different training programs do not directly affect respiratory muscle training effects.

4.2 NRD of respiratory muscle under loudness control

The vocal effort produced significantly greater subglottic pressure during maximum-effort speech (Rosenthal et al., 2014), which may have increased the load during the expiratory phase. However, not all expiratory muscles were significantly associated with loudness. In the loudness control task, the lower ribcage embodied vocalization preferentially because only the 7thIC showed differences while PARA and RA did not. The lack of coordination of abdominal vocalization during the vocalization state in patients with chronic obstructive pulmonary disease was unexpected. We observed a more generalized chest breathing habit in patients with COPD, which may explain the lack of pronounced abdominal muscle vocalization. Some research suggests that SCM muscles may help stabilize human vocalizations (Van Houtte et al., 2013). Pettersen et al. (2005) showed that when healthy ordinary people (professional or non-professional singers) participated in vocalization training and performed vocalization content with different loudness and pitch levels, the activities of the sternocleidomastoid muscle and trapezius muscle increased; this effect was more noticeable when the respiratory demand was strong. Similarly, our observational research showed that the NRD of respiratory muscle was not significantly higher than that of low-loudness vocalizations in SCM in patients with COPD.

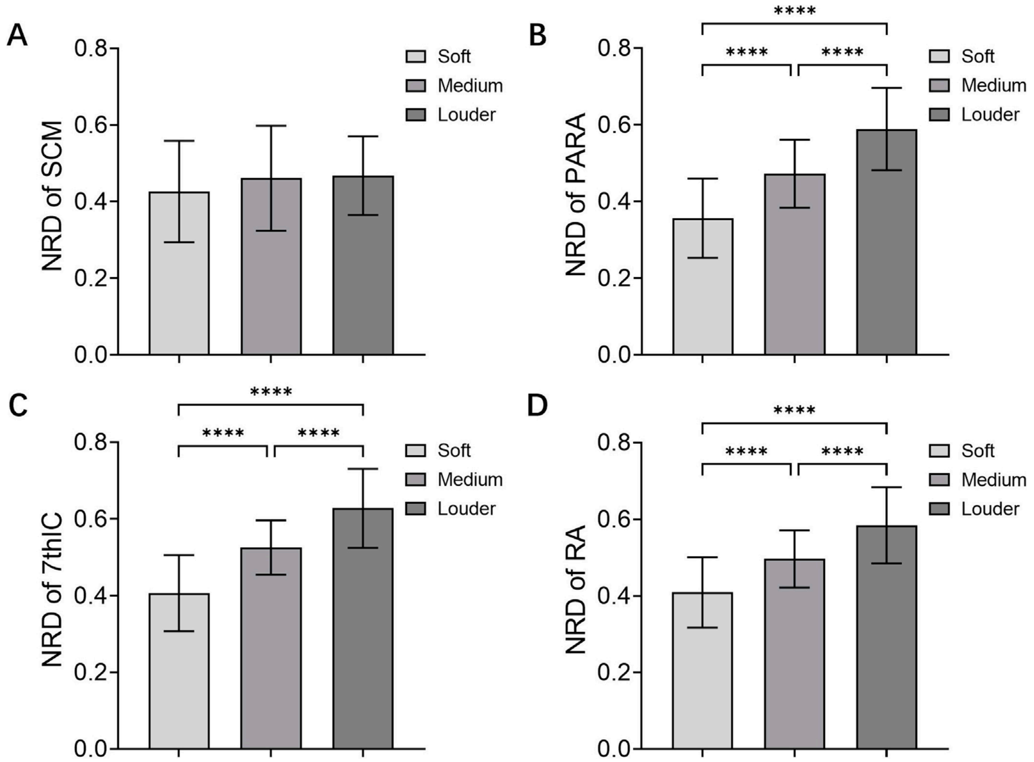

FIGURE 2
NRD of Muscles in loudness control. NRD of **(A)** SCM, **(B)** PARA, **(C)** 7thIC, and **(D)** RA during soft and louder loudness. Abbreviations: NRD, neural respiratory drive; SCM, sternocleidomastoid; PARA, parasternal intercostal muscle; 7thIC, seventh intercostal muscle; RA, rectus abdominis;****$P < 0.001$.

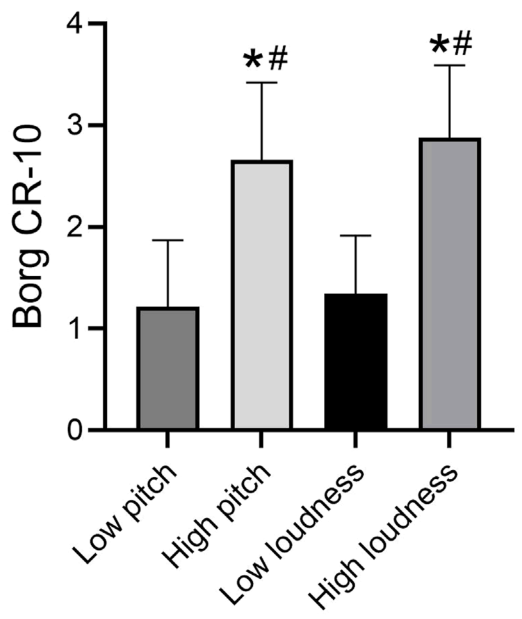

FIGURE 3
Borg breathlessness score in different tasks. Note:*compare with low loudness $P < 0.05$, #compare with High loudness $P < 0.05$.

4.3 Clinical implications

Previous studies have shown that comprehensive pulmonary rehabilitation, based on standard protocols, can improve dyspnea and quality of life in patients (Troosters et al., 2023). However, there are still inconsistencies in vocalization training programs for COPD patients (Fang et al., 2022). This study offers clinical implications for the implementation of vocalization training. Regarding the physiological effects on patients, Fu et al. showed that collective singing and vocal training improved the maximum expiratory pressure and exercise ability of older people in the community (Fu et al., 2018). Lord et al. (2012) reported that the objective physiological state of patients with COPD did not change after vocalization training. However, the content and intensity of singing are not disclosed in vocalization studies, which may be the reason for the differences in the outcome indicators (Patel et al., 2022; Selickman and Marini, 2022). In addition, correlation analysis showed that vocalization of the same loudness produced great stimulation in patients with low respiratory muscle strength. This finding is consistent with the theory that the intercostal and abdominal muscles provide vocal expiratory support (Pettersen et al., 2005; De Troyer and Boriek, 2011). Muscle control and improvement are required to ensure an intensity threshold for training; according to the results of this study, vocalization at

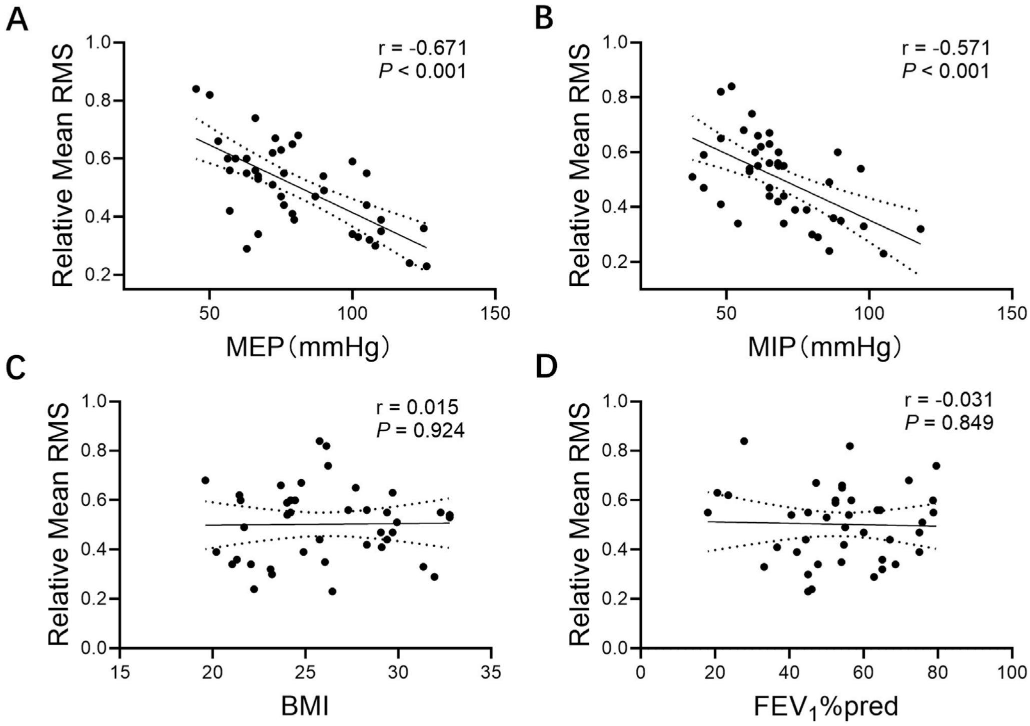

FIGURE 4
Correlations between relative mean RMS and **(A)** MEP, **(B)** MIP, **(C)** BMI, and **(D)** FEV1%pred. Note: Relative mean RMS, mean respiratory muscle RMS% MVC of parasternal intercostal muscle, seventh intercostal muscle, rectus abdominis; MEP, maximal expiratory pressure; MIP, maximal inspiratory pressure; BMI, body mass index; FEV_1%pred, forced expiratory volume in one second % predicted.

a lower intensity may not be sufficient to engage the respiratory muscles fully. At the same time, the degree of dyspnea after the high-loudness task was higher than that after the high-pitched task, while the degree of dyspnea between the low-loudness task and the low-pitched task was not significantly different. Hence, home oxygen therapy, breathing exercises, and other treatments that can improve dyspnea can be combined with vocalization training for improved clinical outcomes. Personalized vocalization prescriptions for patients with COPD should consider the respiratory muscle state of patients rather than simply pursuing vocalization loudness.

4.4 Limitation

Although the study examined the NRD of respiratory muscles to different vocalization tasks, this study still has some limitations. First, this study only addressed the physiological effects of vocal tasks and did not address the psychological effects of vocal training as an artistic engagement. Second, all syllable pitches and loudness are difficult to study because of the patients' limited tolerance. Relatively simple /a/, /i/, and/u/were chosen for this study, considering vocal teachers' suggestions and previous studies. Finally, participants included patients with COPD only who had performed primary vocal exercises to study the effect of training on uninitiated patients. Therefore, the results of the article cannot be generalized to patients who have received full vocalization training.

5 Conclusion

In vocalization training for patients with COPD, focusing on increasing loudness, rather than pitch or vowel, led to the activation of the expiratory muscles. High pitches and loudness produced dyspnea compared with low pitches and low loudness. The maximal expiratory/inspiratory pressure was negatively correlated with respiratory muscle NRD in the same loudness vocalization task.

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 humans were approved by Qingdao Municipal Hospital Ethics Committee. The studies were conducted in accordance with the local legislation and institutional requirements. The 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

ZQ: Data curation, Methodology, Writing–original draft, Writing–review and editing. ZK: Formal Analysis, Methodology, Writing–original draft. JZ: Methodology, Writing–review and editing. DaL: Methodology, Writing–original draft. DoL: Writing–review and editing. XC: Writing–review and editing. KL: Funding acquisition, Methodology, Supervision, Writing–original draft, Writing–review and editing.

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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 of gluteal muscle central activation in individuals with and without patellofemoral pain

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Patellofemoral pain (PFP) is often linked to knee valgus during weight-bearing activities, commonly attributed to gluteal muscle weakness. However, recent research suggests that central nervous system adaptations may also influence muscle function and movement patterns in individuals with PFP. This study compared the central activation ratio (CAR) of the gluteus medius and gluteus maximus between individuals with and without PFP, and assessed the associations between gluteal CAR, frontal plane projection angle of the trunk and lower extremity, and knee function. Twelve individuals without PFP and 10 individuals with PFP participated. We tested CAR of the gluteal muscles with a superimposed burst protocol during a maximum voluntary isometric contraction and evaluated frontal plane kinematics of the trunk and lower extremities during five single leg tasks. Participants with PFP also completed the Anterior Knee Pain Scale (AKPS). Independent t-tests compared CAR between groups, and Pearson correlation coefficients evaluated the associations between CAR, frontal plane kinematics, and AKPS. Individuals with PFP tended to have lower gluteus maximus CAR, though the difference was not statistically significant (PFP: 90.8% $\pm$ 7.0%, Control: 94.4% $\pm$ 3.0%; $p = 0.067$). CAR of both the gluteus maximus ($R = 0.790$, $p = 0.003$) and gluteus medius ($R = 0.584$, $p = 0.038$) were significantly correlated with AKPS scores, and gluteus maximus CAR was associated with trunk lean angle during single leg landing ($R = 0.533$, $p = 0.006$). Our data suggest that higher gluteal CAR is associated with better function in individuals with PFP. Lower gluteus maximus CAR contributes to ipsilateral trunk lean during single leg landing, potentially to reduce external hip moments and muscle demand.

KEYWORDS

patellofemoral pain, central activation, gluteal muscles, frontal plane kinematics, superimposed burst technique

1 Introduction

Research on patellofemoral pain (PFP) has highlighted its association with dynamic knee valgus, a movement impairment characterized by a combination of femoral internal rotation and adduction, knee abduction, tibial external rotation, and ankle pronation during weight-bearing activities (Wilczyński et al., 2020). Such altered movement patterns can result in lateral tracking of the patella and increased loading on the lateral patella, which is thought to contribute to PFP (Powers et al., 2017; Willy et al., 2019; Wilczyński et al., 2020). Given that weakness in the hip muscles is believed to contribute to dynamic knee valgus, physical therapy interventions for PFP often emphasize strengthening the hip abductors and external rotators (Powers et al., 2017; Willy et al., 2019). However, while hip muscle strengthening programs are commonly prescribed, and may successfully reduce pain and hip muscle weakness in the short term, evidence suggests that these interventions may lack efficacy in maintaining long-term improvements and may not adequately correct faulty movement patterns during weight-bearing activities (Willy et al., 2019). This suggests that factors beyond hip muscle strength alone may contribute to the persistence of movement deficits in individuals with PFP.

Central nervous system adaptations in individuals with PFP have been shown in more recent literature, including cortical reorganization of the primary motor cortex, altered spinal reflex excitability, and inhibition of the quadriceps muscles (On et al., 2004; Rio et al., 2016; de Oliveira Silva et al., 2017; Te et al., 2017; Pazzinatto et al., 2019; Liang et al., 2021; Ho et al., 2022; Waiteman et al., 2022). The altered neural pathways observed in persons with PFP may impair their ability to voluntarily activate the affected muscles, which can be crucial for generating the necessary forces to maintain proper joint mechanics and movement patterns during weight-bearing tasks. However, current interventions primarily focus on strengthening weakened muscles rather than directly addressing the underlying altered neural pathways (Bolgla et al., 2016; Willy et al., 2019). This approach might explain the limited long-term effectiveness of conventional rehabilitation for PFP. The impairment in voluntary activation of the gluteal muscles may help explain why hip strengthening protocols alone do not consistently improve muscle strength or correct dynamic knee valgus during weight-bearing activities (Bolgla et al., 2016; Willy et al., 2019). For instance, gluteal muscle inhibition, characterized by the nervous system's inability to fully activate gluteal muscles, may explain the limited success of strengthening interventions (Glaviano et al., 2019; Glaviano and Norte, 2022; Ho et al., 2022). Addressing the underlying neural adaptations, in addition to muscular strengthening, may be required to effectively restore proper joint mechanics and alleviate symptoms.

Voluntary muscle activation can be experimentally quantified using the central activation ratio (CAR) through the superimposed burst (SIB) technique (Glaviano et al., 2019; Glaviano and Norte, 2022). SIB involves applying an exogenous electrical stimulus percutaneously following a maximum voluntary isometric contraction (MVIC). This method allows for the determination of CAR, which represents the ratio of volitionally activated motor units to the total available motor units within a specific muscle (Glaviano et al., 2019; Glaviano and Norte, 2022). By utilizing the SIB technique during MVIC, researchers can assess the extent of muscle activation and identify potential deficits in neuromuscular function.

Recent findings suggest that diminished central activation of the gluteus medius in females with PFP is associated with increased hip adduction during single leg squat and fear-avoidance beliefs (Glaviano and Norte, 2022). However, this study did not include a control group without PFP (Glaviano and Norte, 2022). In addition, an unpublished study (Samuel, 2022) reported that females with PFP had a reduced CAR of the gluteus medius compared to pain-free controls, though no associations were found with knee valgus angle during forward step-down. Notably, the unpublished reports by Samuel focused solely on the gluteus medius, leaving the gluteus maximus unexamined, and assessed CAR in relation to knee valgus during only one weight-bearing task. Females with PFP frequently exhibit weakness in both the gluteus medius and gluteus maximus muscles compared to pain-free individuals, as demonstrated by isometric and isokinetic strength assessments (Van Cant et al., 2014). This finding suggests that hip muscle weakness is not limited to a specific type of muscle contraction or a single gluteal muscle group. Individuals with PFP also demonstrate deficits across multiple aspects of muscle performance, including hip abductor rate of force development, hip muscle power, and dynamic hip strength, such as the force generated during repetition maximum tests (Nunes et al., 2018; Nunes et al., 2019). A wide range of gluteal muscle CAR in females with PFP have also been reported (Glaviano and Norte, 2022), potentially suggesting variability in neuromuscular responses to injury or pain, with a subset experiencing gluteal muscle inhibition. The current literature lacks comprehensive research comparing central activation of both gluteus maximus and gluteus medius muscles between individuals with and without PFP, and exploring the associations between gluteal muscle activation, frontal plane kinematics of the trunk and lower limbs during various weight-bearing tasks, and related functional outcomes. This knowledge gap highlights the need for further studies to better understand the neural mechanisms driving PFP and their impact on movement patterns and functional performance.

Therefore, the primary aim of this study was to compare the CAR of the gluteus medius and gluteus maximus between individuals with and without PFP. A secondary aim was to examine the relationships between CAR of the gluteal muscles and the frontal plane projection angle (FPPA) of the trunk and lower extremities during five weight-bearing activities and patellofemoral joint function. We hypothesized that individuals with PFP would exhibit a lower CAR in both the gluteus medius and gluteus maximus compared to those without PFP. Additionally, we hypothesized that lower gluteal CAR would correlate with altered FPPA of the trunk and lower extremities and reduced function.

2 Materials and methods

2.1 Participants

A sample size calculation was performed prior to the start of the study, indicating that 9 participants per group (PFP and non-PFP) were required to achieve 80% statistical power with a Type I error of 0.05. This estimation was based on prior research examining gluteus medius CAR and sought to detect a potential 8% difference between the two groups (Glaviano and Norte, 2022; Samuel, 2022). Therefore, we aimed to recruit at least 9 participants with PFP and 9 pain-free controls.

The inclusion criteria for the PFP group involved individuals aged 18–45 years who experienced PFP, specifically peri- or retro-patellar pain lasting at least 3 months. A physical examination was conducted to ensure that participants′ pain was not caused by other sources. This examination included palpation around the patellofemoral joint and a patellar compression test, where the patella was pressed while the knee was extended. If the pain did not originate from the patellofemoral joint, participants were excluded from the study (Ho et al., 2021; Ho et al., 2024). For the control group, participants aged 18–45 years who did not have PFP were included. Exclusion criteria for both PFP and control groups included those with a history of traumatic patellar dislocation, previous knee surgeries, or pregnancy.

The Institutional Review Board at the University of Nevada, Las Vegas approved this study (protocol# UNLV-2023-86). Recruitment occurred between 2023 and 2024 via local physical therapy clinics, the university, and other local organizations around Las Vegas. All participants who met the inclusion criteria and consented to the study received detailed information regarding the procedures, risks, and benefits before participating.

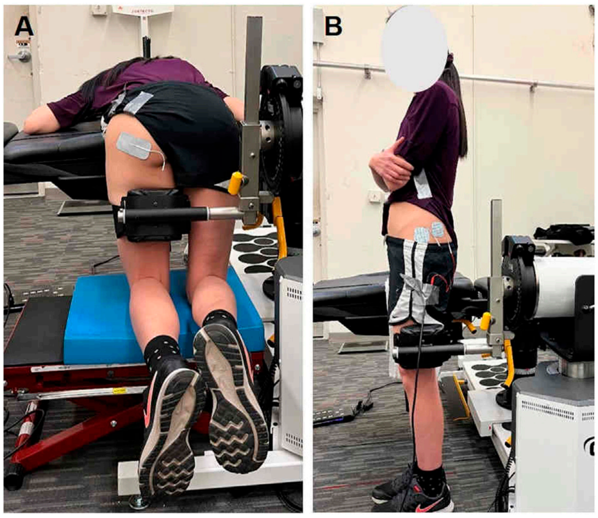

FIGURE 1
Participant positioning and electrode configurations for central activation ratio (CAR) assessments of the **(A)** gluteus maximus and **(B)** gluteus medius.

2.2 Procedures

All participants completed two examination sessions on the same day: the first assessed frontal plane kinematics during weight-bearing activities, followed by an evaluation of central activation of the gluteal muscles. In participants with PFP, the assessment was performed on the symptomatic or more symptomatic limb. For those in the control group, the dominant leg was studied, which was identified as the preferred limb used for kicking a ball (van Melick et al., 2017).

2.2.1 Pain and function assessment

Before assessing frontal plane kinematics during weight-bearing activities, participants with PFP were asked to report their pain and functional status using validated self-report measures. The Numeric Pain Rating Scale (NPRS) was employed to evaluate their usual pain during daily activities, with 0 representing no pain and 10 representing extreme pain. NPRS has been shown to be valid, reliable and appropriate for use in clinical practice (Williamson and Hoggart, 2005). To assess patellofemoral joint function, the Anterior Knee Pain Scale (AKPS) was administered in participants with PFP. The AKPS is a reliable self-report tool designed specifically for evaluating function in individuals with PFP. It includes 13 weighted questions, with a total score of 100 indicating no disability. Higher scores reflect better functional status (Crossley et al., 2002). All participants were also asked to report their physical activity levels using the Global Physical Activity Questionnaire (GPAQ), developed by the World Health Organization to assess physical activity and sedentary behaviors. The GPAQ consists of 16 questions covering three domains: work-related activity, travel, and recreational activities. It has been validated as an effective tool for measuring moderate to vigorous physical activity and shows a strong correlation with the International Physical Activity Questionnaire (Bull et al., 2019).

2.2.2 Assessment of frontal plane kinematics during weight-bearing activities

To capture frontal plane kinematics of the trunk and lower extremities for each participant, spherical matte stickers were used as markers. These markers were placed on the manubrium of the sternum, bilateral anterior superior iliac spines (ASIS), bilateral patellae, and bilateral mid-talus. Participants were then videotaped performing a series of weight-bearing tasks: single leg squat, single leg hop, single leg landing, forward step-down, and lateral step-down. All movements were recorded using an iPhone 13 Pro Max camera (Apple Inc., Cupertino, CA, United States) set to 30 frames per second in high definition with a 1x zoom. For all participants, the camera was mounted on a tripod positioned 0.71 m above the ground and 3.1 m in front of the participants to ensure consistent two-dimensional (2D) frontal plane measurements.

For the single leg squat test, participants stood on their testing limb, performed a squat, flexing their affect knee until they reached their maximal knee flexion without loss of balance. During the single leg hop test, they were instructed to hop as far as possible on the testing leg (Ho et al., 2019). In the single leg landing task, participants started on their testing leg atop a 6 inch platform, stepped forward, and landed on the same leg. The forward step-down test required participants to stand on their testing leg on a 30 cm box, lower the other leg forward to touch the ground, and then return to the starting position within a 2-second period (Lee and Powers, 2014). For the lateral step-down test, participants stood on their testing leg on a 30 cm box, lowered the other leg to the side to touch the ground, and returned to the starting position within 2 seconds (Ho et al., 2024). Each task was performed three times, with 3-minute rest intervals between tasks. Additional breaks were allowed upon request.

2.2.3 Assessment of central activation of the gluteal muscles

We assessed the CAR of the gluteal muscles using the SIB technique (Glaviano et al., 2019; Glaviano and Norte, 2022). Specifically, the gluteus maximus was examined while participants lie prone with 90 degrees of hip flexion and 90 degrees of knee flexion, and performed a MVIC by extending the hip against the dynamometer attachment arm. To assess CAR in the gluteus maximus, two adhesive electrodes were placed just inferior to the posterior gluteal line of the ilium and medial to

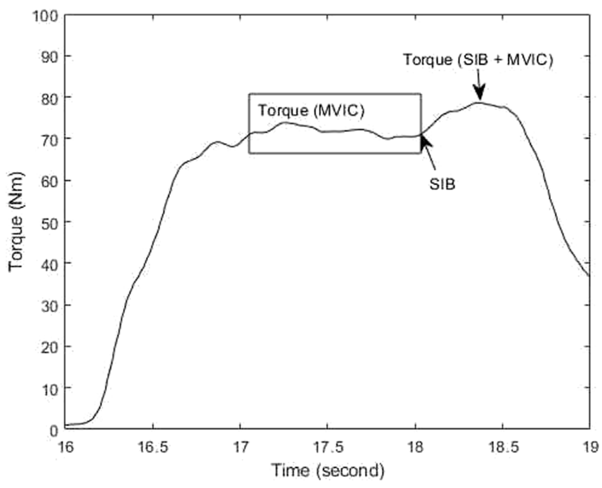

FIGURE 2
Central activation ratio (CAR) calculation during the superimposed burst (SIB) technique.

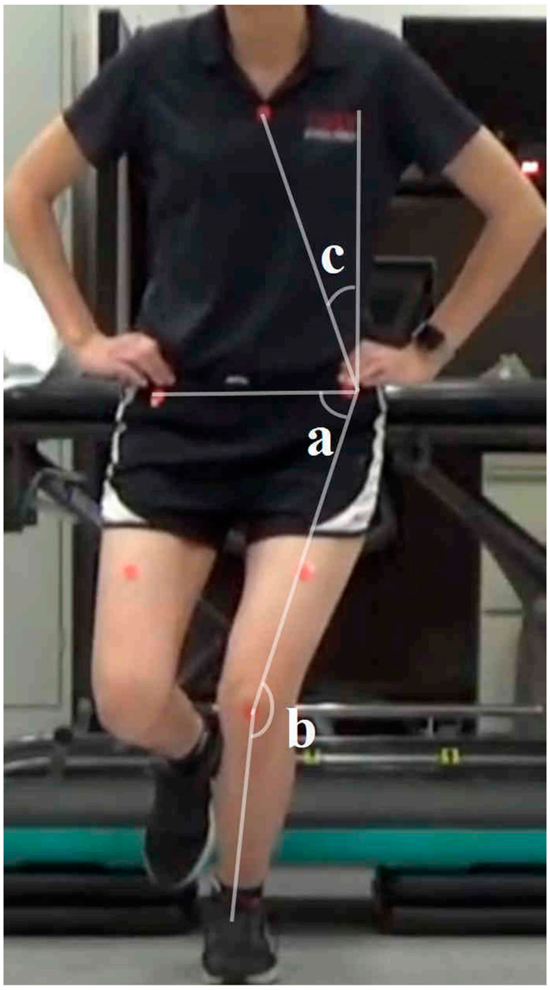

FIGURE 3
Two-dimensional frontal plane kinematics measured during single leg squat. TLA = c; hip FPPA = 90 degrees - a; knee FPPA = 180 degrees - b; DVI = a + b. Abbreviations: FPPA = frontal plane projection angle; DVI = dynamic valgus index; TLA = trunk lean angle.

the greater trochanter along the line of its insertion to the iliotibial band (Figure 1A). For assessment of CAR in the gluteus medius, participants stood on their non-test limb, and performed MVIC by abducting their test hip into the dynamometer attachment arm. The electrodes for the gluteus medius were placed at the area superior to the greater trochanter and the central area of the most superior aspect of the muscle (Figure 1B) (Te et al., 2017; Glaviano et al., 2019).

Prior to recording, a practice MVIC trial was performed, followed by a 2-minute rest period. The torque generated by the muscles and electrical stimulation was recorded by a motored dynamometer (Humac Norm, Computer Sports Medicine Inc., Stoughton, MA, United States). Participants were instructed to gradually increase their torque level to reach their MVIC within a span of 2 seconds. Once the torque reached a plateau during MVIC, a manually triggered brief electrical stimulus was delivered to the target muscle using a biphasic constant current stimulator (DS8R, Digitimer®, Welwyn Garden City, United Kingdom) and train generator (DS2A, Digitimer®, Welwyn Garden City, United Kingdom). The electrical stimulus consisted of a 100 ms train of 10 square-wave pulses of 600 us pulse width at 100 Hz, aiming to increase the torque output by recruiting the un-recruited motor units (Norte et al., 2015; Gilfeather et al., 2019; Glaviano and Norte, 2022). Manually triggering of electrical stimulation was chosen to reduce the number of MVIC trials required for CAR assessment, and to reduce possible fatigue of the participants. All participants received verbal encouragement during MVIC. To minimize fatigue, each participant performed at least two trials but no more than five MVIC trials, with a 2-minute rest period between trials for each muscle group. Additional rest was provided upon request.

2.3 Data analysis

2.3.1 Calculation of gluteal central activation

The central activation profile of hip muscles was analyzed using a custom MATLAB code (MathWorks. MATLAB. Natick, MA, United States), which involved the calculation of the CAR using the equation below (Glaviano et al., 2019; Glaviano and Norte, 2022). The torque during MVIC was defined as the torque output during the 100-millisecond window prior to the electrical stimulation, while the torque during SIB and MVIC was defined as the highest torque generated after the delivery of the electrical stimulation (as shown in Figure 2) (Glaviano et al., 2019; Glaviano and Norte, 2022). The CAR values for the gluteus medius and gluteus maximus from the collected trials were averaged and used for statistical analyses.

$$CAR = \frac{Torque\ (MVIC)}{Torque\ (SIB + MVIC)} \times 100\%$$

2.3.2 Calculations of frontal plane kinematics during weight-bearing activities

The videos of participants were analyzed using Kinovea software (version 0.8.15, Kinovea, Bordeaux, France). Four primary measures were extracted from the video footage: trunk lean angle (TLA), knee frontal plane projection angle (FPPA), hip FPPA, and dynamic valgus index (DVI). These angles were measured at the peak of knee flexion (Scholtes and Salsich, 2017; Ho et al., 2024). The measurement process began by drawing a vertical reference line extending upward from the ipsilateral ASIS. To define the pelvic segment, a line was drawn connecting the markers on the bilateral ASIS landmarks. The thigh segment was represented by a line from the midpoint of the patella to the ipsilateral ASIS, while the shank segment was defined by a line from the midpoint of the patella to the midpoint of the ankle (Ho et al., 2024).

The TLA was calculated as the angle between the vertical reference line and the line connecting the ipsilateral ASIS to the sternal marker (Dingenen et al., 2014). A lower TLA indicates a greater lean of the trunk towards the testing limb. The knee FPPA was determined by subtracting the angle between the thigh and shank segments from 180°. A higher knee FPPA reflects greater knee valgus in the testing limb (Scholtes and Salsich, 2017). The hip FPPA was calculated by subtracting the angle between the pelvic and thigh segments from 90°. A higher hip FPPA signifies increased hip adduction of the testing limb (Scholtes and Salsich, 2017). The DVI, which accounts for both knee valgus and hip adduction, was computed as the sum of knee FPPA and hip FPPA (Scholtes and Salsich, 2017) (Figure 3). A higher DVI indicates a greater overall degree of knee valgus and/or hip adduction. For each task, these four angles were measured from three repetitions, and the average value for each task was used in the statistical analysis.

2.4 Statistical analysis

The primary outcome measures included the CAR of the gluteus maximus and gluteus medius, as well as the frontal plane kinematics during five weight-bearing tasks: single leg squat, single leg hop, single leg landing, forward step-down, and lateral step-down. Independent t-tests were conducted to compare CAR measurements of the gluteus maximus and gluteus medius between the control and PFP groups. Pearson correlation coefficients were used to assess the relationships between the CAR of the gluteus maximus and gluteus medius and the frontal plane kinematics of the trunk and lower extremities (including knee FPPA, hip FPPA, DVI, and TLA) across all participants during the five weight-bearing tasks. Additionally, the correlations between gluteal CARs and the AKPS were examined for individuals with PFP using Pearson correlation coefficients. Correlations were classified as follows: small (0.1–0.3), moderate (0.3–0.5), large (0.5–0.7), very large (0.7–0.9), and extremely large (greater than 0.9) (Hopkins et al., 2009). All statistical analyses were conducted using the Statistical Package for the Social Sciences (version 23.0; IBM Corporation, Armonk, NY). A significance level was established with a threshold of $p \leq 0.05$.

3 Results

3.1 Participant characteristics

The procedure was successfully completed with 10 participants experiencing PFP and 12 pain-free controls. Independent t-tests indicated that there were no differences between groups in age, body mass index (BMI) and activity level. The chi-square test confirmed that the sex distribution was similar between the two groups (Table 1).

3.2 Gluteal central activation

Independent t-tests revealed no statistically significant difference in the CAR of the gluteus maximus between the PFP group (90.8% ± 7.0%) and the control group (94.4% ± 7.0%) ($p = 0.067$), despite an observed 3.6% decrease in CAR in the PFP group. This lack of statistical significance is likely attributable to the large standard deviation observed in the gluteus maximus CAR within the PFP group, suggesting substantial variability in central activation levels among individuals. Similarly, there was no statistically significant difference in the CAR of the gluteus medius between the PFP group (93.3% ± 4.7%) and the control group (95.2% ± 3.7%) ($p = 0.151$) (Figure 4). Therefore, our hypothesis that individuals with PFP would exhibit lower CAR values in both the gluteus maximus and gluteus medius compared to controls was not fully supported. While individuals with PFP tended to have lower CAR in the gluteus maximus, it did not reach statistical significance.

3.3 Correlations between gluteal central activation and kinematics/ functional outcomes

Pearson correlation coefficient analysis revealed a statistically significant large correlation between the CAR of the gluteus maximus and the TLA during single leg landing across all participants ($R = 0.533$, $p = 0.006$; Table 2). No significant associations were observed between gluteal CAR values and the kinematics of other tasks. Additionally, there was a significant correlation between the CAR of the gluteus maximus and scores on the AKPS in individuals with PFP ($R = 0.790$, $p = 0.003$; Table 2). A significant correlation was also found between the CAR of the gluteus medius and AKPS scores ($R = 0.584$, $p = 0.038$; Table 2). Consequently, our hypothesis that lower gluteal CAR would correlate with altered FPPA of the trunk and lower extremities, as well as reduced function, was partially supported. Specifically, we observed a relationship between gluteus maximus CAR and TLA during single leg landing, as well as between gluteal CAR and AKPS scores.

4 Discussion

Our primary aim was to compare the CAR of the gluteus medius and gluteus maximus between individuals with and without PFP. A secondary objective was to examine the relationship between CAR of the gluteal muscles and the frontal plane kinematics of the trunk and lower extremities during five weight-bearing activities and patellofemoral joint function. Our hypotheses were partially supported. We observed a 3.6% decrease in gluteus maximus CAR in individuals with PFP compared to controls, but the difference did not reach statistical significance. Gluteus medius

TABLE 1 Participant characteristics of the patellofemoral pain (PFP) and control groups.

	PFP	Controls	P Value
Age (years)	22.6 ± 2.8	24.2 ± 1.8	0.052
Sex	4 males; 6 females	4 males; 8 females	0.746
Body Mass Index (BMI) (kg/m^2)	23.5 ± 3.1	24.0 ± 4.2	0.419
Physical activity level (MET. min/week)	4,205 ± 2,575	2,510 ± 1,732	0.160
Anterior Knee Pain Scale	81.4 ± 9.3	N/A	N/A
Duration of symptoms (months)	35.4 ± 27.2	N/A	N/A
Average pain	2.9 ± 1.7	N/A	N/A

Data is presented as mean ± standard deviation.

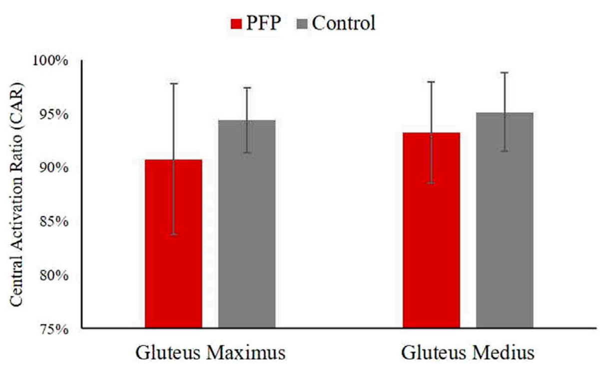

FIGURE 4
Comparisons of central activation ratio of the gluteus maximus and gluteus medius. Data is presented as mean ± standard deviation. PFP, patellofemoral pain.

CAR was not different between groups. With respect to our secondary aim, higher CAR values for gluteus maximus and gluteus medius were associated with higher AKPS scores. Additionally, reduced gluteus maximus CAR was linked to lower TLA, reflecting greater ipsilateral trunk lean toward the standing limb during movement. This may represent a compensatory strategy to decrease external hip moments and reduce the demand on hip muscles during single leg landing.

The current literature demonstrates considerable variability in CAR values among individuals with and without PFP. In asymptomatic individuals, Gilfeather et al. (2019) reported CAR values of 96.4% for the gluteus medius and 86.9% for the gluteus maximus, while an unpublished study (Samuel, 2022) reported an even higher gluteus medius CAR of 98.4% in asymptomatic individuals. In comparison, in our study, our observed gluteus medius CAR of 95.15% in pain-free controls aligned with previous research, whereas our observed gluteus maximus CAR of 94.40% was higher than previously reported values in pain-free groups. Among those with PFP, Glaviano and Norte (Glaviano and Norte, 2022) found CAR values of 90.5% for the gluteus medius and 84.0% for the gluteus maximus in females with PFP, whereas Samuel (Samuel, 2022) reported a higher gluteus medius CAR of 95.9% in females with PFP. Our study's CAR values for the PFP group (gluteus maximus = 90.8 ± 7.0%; gluteus medius = 93.3 ± 4.7%) fell within these reported ranges. Importantly, consistent with prior research (Glaviano and Norte, 2022), we also observed higher CAR for gluteus medius compared to gluteus maximus in persons with PFP. This disparity in hip muscle activation may contribute to increased knee valgus in persons with PFP, as insufficient strength or neuromuscular control in the hip abductors could impair their ability to abduct the thigh during weight-bearing activities, leading to altered movement patterns.

In comparing gluteal CAR between individuals with and without PFP, unpublished data by Samuel (Samuel, 2022) reported a 2.5% reduction in the CAR of the gluteus medius among those with PFP (PFP = 95.9%; control = 98.4%). Our study expanded on this by including an analysis of the gluteus maximus. Although the difference in CAR for the gluteus maximus and gluteus medius between individuals with and without PFP did not reach a statistical significance in our study, comparisons for the gluteus maximus approached significance, indicating a possible lower gluteus maximus CAR in participants with PFP compared to pain-free controls ($p = 0.067$). Despite reaching our targeted sample size based on power estimates, we did not observe a statistically significant difference between groups. A key factor contributing to this outcome was likely the wide variability in gluteal CAR values among participants, with gluteus maximus CAR ranging from 80.5% to 99.7%, and gluteus medius CAR from 84.4% to 98.3%. These observations suggest the potential presence of subgroups within the PFP population, where some individuals may not exhibit central activation deficits despite having PFP. Therefore, identifying these subgroups within the PFP population is crucial. Since not all individuals with PFP may present with diminished gluteal CAR, further research is needed to identify subgroups of PFP patients with central activation deficits, explore how these deficits relate to movement impairments, and establish a threshold for defining reduced gluteal CAR.

A key finding from our study was the correlation between lower CAR values and increased ipsilateral trunk lean during the single leg landing task. This suggests that participants may lean toward the stance limb to reduce external hip moments, possibly as a compensatory strategy to offload the demand on the hip muscles. Glaviano and Norte (Glaviano and Norte, 2022), using three-dimensional (3D) motion analysis, reported an association between lower gluteus maximus CAR values and greater hip adduction during single leg squat. However, their study did not examine trunk lean, focusing solely on hip kinematics during the squat. The difference in tasks and methodological approaches may

TABLE 2 Correlations between central activation ratio (CAR) of gluteal muscles, frontal plane kinematics and function.

	CAR of gluteus maximus		CAR of gluteus medius	
	R	P value	R	P value
Single Leg Squat				
Trunk Lean Angle	0.042	0.428	0.283	0.101
Hip FPPA	−0.201	0.191	0.206	0.179
Knee FPPA	−0.214	0.176	−0.164	0.233
Dynamic Valgus Index	−0.223	0.166	−0.010	0.482
Single Leg Hopping				
Trunk Lean Angle	0.228	0.160	0.007	0.487
Hip FPPA	−0.122	0.299	0.003	0.495
Knee FPPA	0.006	0.490	0.017	0.471
Dynamic Valgus Index	−0.066	0.389	0.010	0.482
Single Leg Landing				
Trunk Lean Angle	0.533	0.006*	0.305	0.083
Hip FPPA	−0.018	0.469	0.039	0.432
Knee FPPA	0.082	0.362	0.150	0.253
Dynamic Valgus Index	0.040	0.432	0.109	0.314
Forward Step Down				
Trunk Lean Angle	0.252	0.135	0.194	0.194
Hip FPPA	−0.04	0.494	0.022	0.461
Knee FPPA	0.40	0.432	−0.008	0.486
Dynamic Valgus Index	0.023	0.461	0.006	0.490
Lateral Step Down				
Trunk Lean Angle	0.080	0.365	0.248	0.132
Hip FPPA	−0.167	0.235	0.138	0.270
Knee FPPA	−0.003	0.495	0.120	0.298
Dynamic Valgus Index	−0.087	0.353	0.143	0.262
Anterior Knee Pain Scale	0.790	0.003*	0.584	0.038*

Abbreviations: CAR, central activation ratio; FPPA, frontal plane projection angle.
* Indicates a statistically significant difference using a Pearson correlation coefficient analysis.

explain the divergence in findings between our study and theirs. Specifically, the 3D motion analysis employed by Glaviano and Norte offers enhanced precision in capturing complex movement patterns, potentially explaining why they observed relationships with hip adduction that were not evident in our data. When comparing our observations with Samuel's unpublished study, which utilized a similar 2D motion analysis approach for the lower extremities during a forward step-down task, neither study identified a significant relationship between 2D frontal plane kinematics of the lower extremities and gluteal CAR values. These discrepancies across studies highlight the complexity of the relationships between gluteal CAR and movement kinematics. This suggests the need for future research utilizing 3D motion analysis to investigate how gluteal activation patterns influence kinematics of both the lower extremity and trunk in persons with PFP. Examining a wider range of weight-bearing tasks, including those that challenge frontal and transverse plane stability, could offer deeper insights into compensatory movement strategies and their implications for individuals with PFP. Additionally, future studies should aim to explore the potential influence of individual variability, such as differences in pain severity, activity levels, or neuromuscular adaptations, which may contribute to the observed variability in gluteal CAR and its relationship with kinematics.

Our study found that lower CAR in both the gluteus maximus and gluteus medius were linked to decreased function, as measured by the AKPS. These results highlight the crucial role of central activation in the gluteal muscles for maintaining proper patellofemoral joint function in persons with PFP. This is consistent with the findings of Glaviano and Norte (Glaviano and Norte, 2022), who identified a correlation between gluteus medius CAR values and the Fear-Avoidance Belief Questionnaire-Physical Activity scores. Their research suggests that individuals with PFP who have lower gluteus medius CAR values are more likely to experience greater fear avoidance toward physical activity. These insights highlight the potential role of gluteal central activation in managing both functional limitations and psychological fear associated with PFP.

A potential intervention to address decreased central activation of the gluteal muscles in persons with PFP is transcranial direct current stimulation (tDCS), a non-invasive brain stimulation technique that delivers a direct weak electric current to the brain, and can modulate cortical excitability (Nitsche and Paulus, 2000; Lefaucheur et al., 2017), improve motor function (Hummel et al., 2005; Liang et al., 2020), or alleviate pain (Yang et al., 2024). The modulatory effect is influenced by the positioning and polarity of the scalp electrodes. Anodal stimulation enhances cortical excitability, while cathodal stimulation reduces it, and bimodal stimulation simultaneously increases excitability in the region beneath the anode and decreases excitability in the region beneath the cathode (Nitsche and Paulus, 2000). Our recent research study has demonstrated the feasibility of using bimodal tDCS to target gluteal corticomotor function in combination with hip muscle strengthening in individuals with PFP (Ho et al., 2024). In our earlier study, we used the bimodal montage, where tDCS was applied with the anode positioned over the primary motor cortex contralateral to the more painful limb and the cathode over the ipsilateral motor cortex, aiming to enhance motor cortex function and optimize its effects. Future research should explore the use of tDCS in individuals with PFP who exhibit reduced gluteal CAR, examining its potential clinical benefits in improving gluteal muscle activation, hip muscle performance, trunk and lower extremity kinematics, and overall functional outcomes.

This study has a few limitations. First, the findings have limited generalizability to the broader population due to the relatively young age of the participants, with a mean age of 22.6 years in the PFP group and 24.2 years in the control group. As such, the results may not fully reflect the variations seen in a broader PFP population that includes individuals of different ages, and caution should be exercised when attempting to apply these findings to individuals outside the specific age range studied. Further research including a wider age range would be necessary to better understand the relationship between gluteal CAR and PFP across different age groups. Additionally, participants in the PFP group exhibited a wide range of physical activity levels, with most being highly active, as indicated by their GPAQ scores. However, no significant difference in overall physical activity was observed between the PFP group and the control group. It is also interesting to note that GPAQ scores were not correlated with gluteal muscle CARs in this study. This may be due to the fact that, while the GPAQ assesses various aspects of physical activity in daily life, it does not specifically target activities that involve hip muscle training.

5 Conclusion

Our study did not reveal significant group differences in gluteal CAR, although individuals with PFP showed a tendency for lower gluteus maximus CAR. In addition, greater gluteal central activation was associated with better function in individuals with PFP. The observed relationship between lower gluteus maximus CAR and ipsilateral trunk lean during single leg landing suggest a compensatory strategy aimed at reducing external hip moments and offloading the demand on hip muscles. This highlights the potential role of neuromuscular adaptations in the movement patterns of individuals with PFP.

Future research should prioritize large-scale studies to better understand the heterogeneity of gluteal central activation within the PFP population. Specifically, efforts should focus on identifying subgroups characterized by diminished gluteal central activation and exploring the clinical implications of such deficits. Additionally, studies should investigate the biomechanical and neuromuscular mechanisms linking gluteal central activation to movement outcomes, with a specific focus on weight-bearing tasks that require dynamic stability. Furthermore, longitudinal research could examine the effects of targeted interventions, such as neuromuscular training, cortical priming using tDCS, or hip strengthening programs, on gluteal central activation and movement biomechanics. Investigating whether these interventions can mitigate abnormal kinematics or improve functional outcomes would provide valuable insights for optimizing rehabilitation strategies for individuals with PFP.

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 humans were approved by Institutional Review Board at the University of Nevada, Las Vegas. The studies were conducted in accordance with the local legislation and institutional requirements. The participants provided their written informed consent to participate in this study.

Author contributions

K-YH: Conceptualization, Data curation, Formal Analysis, Funding acquisition, Investigation, Methodology, Project administration, Resources, Software, Supervision, Validation, Visualization, Writing–original draft, Writing–review and editing. MC: Data curation, Formal Analysis, Funding acquisition, Investigation, Methodology, Project administration, Software, Visualization, Writing–original draft, Writing–review and editing. JD: Data curation, Formal Analysis, Funding acquisition, Investigation, Methodology, Project administration, Software, Visualization, Writing–original draft, Writing–review and editing. JK: Data curation, Formal Analysis, Funding acquisition, Investigation, Methodology, Project administration, Software, Visualization, Writing–original draft, Writing–review and editing. JL: Conceptualization, Funding acquisition, Investigation, Methodology, Project administration, Resources, Supervision, Validation, Writing–original draft, Writing–review and editing.

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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.

Generative AI statement

The authors utilized OpenAI's ChatGPT (version GPT-4) to correct grammatical errors.

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Influence of running speed, inclination, and fatigue on calcaneus angle in female runners

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Running is a popular form of physical activity with significant health benefits, but improper technique can lead to running-related injuries. This study investigates the influence of running speed, incline, and fatigue on calcaneus eversion/inversion angle at heel strike, maximum eversion angle, and range of motion, factors associated with lower limb injuries. Fifteen injury-free female runners participated in this study. Kinematic data were collected using a 3D motion capture system with reflective markers placed directly on the skin through specially modified running shoes. The runners performed treadmill trials at varying speeds (10, 12, and 14 km/h) and inclines (0°, 5°, and 10°), both before and after a fatigue-inducing 30-min run. The results indicate that higher speeds were associated with an increase in inversion angle at heel strike (p = 0.05) and range of motion (p = 0.02 before fatigue), both of which are linked to chronic ankle instability and Achilles tendinopathy. Running at an incline reduced both maximum eversion angle (p = 0.002 after fatigue) and range of motion (p = 0.003 after fatigue), suggesting a protective effect against excessive eversion. Fatigue increased range of motion (p = 0.05), which is a risk factor for instability and overuse injuries. These findings suggest that running at higher speeds and in a fatigued state may increase the likelihood of injuries due to increased range of motion, whereas incline running may mitigate this risk by reducing excessive eversion and range of motion. Understanding these biomechanical changes can inform injury prevention strategies for runners.

KEYWORDS

biomechanics, eversion, inversion, injury prevention, running related injury, 3D kinematics

1 Introduction

Physical activity is essential for promoting health, enhancing wellbeing, and increasing life expectancy by reducing the risk of chronic diseases such as cardiovascular disease, type 2 diabetes, and cancer (Durstine et al., 2013). Among various forms of exercise, running is a highly accessible and cost-effective option that most individuals can engage in without special preparation. Research shows (Pedisic et al., 2020) that running significantly lowers the risk of all-cause mortality (27%), cardiovascular disease (30%), and cancer (23%) when compared to a sedentary lifestyle. However, while running offers considerable health benefits, improper techniques can lead to injuries that affect quality of life and healthcare costs. Therefore, understanding the risk factors associated with these injuries is crucial for mitigating their negative effects (Ceyssens et al., 2019), as such injuries can compel individuals to reduce their physical activity, further burdening healthcare systems.

Running-related injuries (RRIs) are common and include conditions such as runner's knee, Achilles tendonitis, stress fractures, shin splints, muscle strains, ankle sprains, plantar fasciitis and iliotibial band syndrome (Kakouris et al., 2021). Estimates indicate that over 40.2% of RRIs are associated with foot and ankle mechanics, with more than a third linked to abnormal joint movements (Kakouris et al., 2021). One such abnormality that significantly contributes to RRIs is excessive ankle eversion. When the eversion angle is excessively high, it leads to an over-pronation of the foot. In the stance phase, this excessive eversion and foot pronation cause an internal rotation of the tibia, femur, and pelvis. This alters the kinematics of the lower limb, creating compensatory movements in adjacent joints, which can strain the musculoskeletal system and increase the risk of injury. For instance, excessive ankle eversion (Stacoff et al., 2000; Clement et al., 1981; Nigg and Morlock, 1987; van Mechelen, 1992) during mid-stance generates substantial strain on the medial fibers of the Achilles tendon thereby elevating the risk of developing Achilles tendinopathy (Lorimer and Hume, 2014; Ryan et al., 2009). Moreover, excessive eversion during the stance phase can overload critical muscles such as the flexor digitorum brevis, tibialis posterior, and soleus, potentially leading to conditions like medial tibial stress syndrome (Beck and Osternig, 1994; Becker et al., 2018; Franklyn and Oakes, 2015; Reinking et al., 2017). Various studies have explored lower limb kinematics concerning these injuries, often comparing kinematic parameters between injured and non-injured groups (Ryan et al., 2009; Ferber et al., 2010; Grau et al., 2011; Coventry et al., 2006; Latorre-Román et al., 2019; Ye et al., 2024; Fukuchi et al., 2017; Hein and Grau, 2014; Donoghue et al., 2008). Studies by Donoghue et al. (2008) and Ryan et al. (2009), have identified significant differences in eversion angles between these groups, suggesting a possible correlation between eversion and injury risk.

The etiology of running-related injuries (RRIs) is frequently associated with suboptimal conditioning, excessive training loads, or elevated fatigue levels, which collectively compromise the body's capacity to absorb and dissipate impact forces. Variability in loading rates and the presence of high-frequency components within ground reaction forces are critical factors influencing injury risk; specifically, elevated loading rates correlate with increased impact forces, while high-frequency components may signify underlying biomechanical stress (Wang et al., 2023). For instance, loading rates that exceed 80 body weights per second ($BW{\cdot}s^{-1}$) have been associated with a heightened risk of injury among runners, (Yang et al., 2024), while heel strikes are known to generate significant impact forces that, when subjected to repetitive application, may contribute to an increased likelihood of injury over time (Senapati et al., 2024). Jones et al. (2017) On the other hand, several kinematic studies have also shown that changes in kinematics due to fatigue from prolonged running can contribute significantly to the development of RRI (Derrick et al., 2002; Koblbauer et al., 2014) Research by Cheung and Ng (Cheung and Ng, 2007) shows that fatigue significantly alters movement characteristics by limiting physical abilities, making it a critical factor in movement-related injuries. In addition to fatigue, running biomechanics are influenced by different factors such as running speed and incline. As speed increases, joint kinematics and muscle activation patterns change, thereby modifying the biomechanical load on the lower limbs. For instance, faster-running correlates with increased joint flexion and more dynamic movement patterns, affecting the distribution of forces throughout the body (Zandbergen et al., 2023). The duty factor—referring to the proportion of the gait cycle during which the foot is in contact with the ground—decreases at higher speeds, limiting the period available for absorbing impact forces and potentially elevating the risk of injury (Van Hooren et al., 2024). Moreover, Jacobs and Berson (Jacobs and Berson, 1986) found that an increase in training speed is directly correlated with injury. While some other studies found no correlation between training speed and the risk of sustaining injury (van Gent et al., 2007; Rauh et al., 2006). Incline running also significantly affects biomechanics, with uphill running increasing the load on the tibia and placing additional strain on lower leg muscles, raising the risk of tibial stress injuries (Rice et al., 2024). However, incline running has been shown to reduce loading rates and peak vertical ground reaction forces, particularly when speed is adjusted to maintain iso-efficiency, as demonstrated by Williams et al. (2020) Both speed and incline alter spatiotemporal parameters, such as stride length and cadence. Given the substantial influence of these factors on running kinematics, it is crucial to explore how they contribute to the incidence of RRIs. In this context, we aim to monitor the eversion angle of the foot under varying conditions—speed, incline, and fatigue.

Given the complex nature of running biomechanics and the various factors influencing injury risk, it is essential to accurately assess and analyze lower limb kinematics. Kinematic analysis of the lower limbs can be conducted using several methods, with the high-speed dual fluoroscopic imaging system (DFIS) (Ye et al., 2024) and 3D motion capture systems utilizing skeletal markers being among the most accurate (Stacoff et al., 2000). However, these methods are either invasive (Stacoff et al., 2000) or potentially harmful to participants due to X-ray exposure (Ye et al., 2024) making them unsuitable for widespread application. Another slightly less accurate but non-invasive method is the gold standard 3D motion capture with reflective skin markers, allowing for the quantification of angles between body segments such as the calcaneus and tibia (Novacheck, 1998). Several studies have been published using 3D motion capture systems to investigate lower limb kinematics. However, numerous studies have attached reflective markers to the running shoe (Ferber et al., 2010; Fukuchi et al., 2017; Napier et al., 2019), which may lead to inaccurate measurements of subtalar joint movement. To our knowledge, there has been one study that used custom-made shoes that allowed researchers to position markers directly on the skin; however, they investigated the effect of barefoot running (Hein and Grau, 2014).

Present study aims to further explore lower limb kinematics, particularly examining the eversion angle of the ankle joint, which is critical in the development of RRIs. Specifically, we will investigate how different running speeds, inclines, and fatiguing affect the calcaneus angle in injury-free female runners with a heel-to-toe heel strike pattern. By systematically altering running conditions, we seek to identify biomechanical changes that may heighten susceptibility to RRIs.

2 Materials and methods

2.1 Participants

The study involved 15 active female runners with an average age of 28 ± 13 years, an average body mass of 60 ± 6 kg, and an average height of 168 ± 4 cm. Participants were recruited from the Faculty of Sport and local sports clubs as they were willing to participate. They completed a questionnaire to assess their eligibility based on criteria such as level of physical activity, absence of neurological or chronic non-communicable diseases, and fitness requirements relevant to the study. Importantly, the questionnaire included questions on heel strike foot pattern, an important inclusion criterion. All participants confirmed the heel strike pattern, were physically active for at least 5 h per week, had no neurological or chronic non-communicable diseases, and gave written informed consent prior to inclusion in the study. The heel strike foot pattern was verified prior to the measurements by analysing the participants' heel strike patterns with a high-speed camera. The study was approved by the Committee for Ethical Issues in Sports at the University of Ljubljana, Slovenia (1/2023).

2.2 Study design

The participants visited the laboratory once, where an inclusion measurement was first carried out. The main criterion for inclusion in the study was the assumption of a rearfoot running technique. To confirm this, each participant's heel strike pattern was analysed using a high-speed camera before measurements began. Participants performed short treadmill runs at speeds of 10 and 14 km/h with and without incline.

After the inclusion measurements, reflective markers were attached to the participants, who wore custom running shoes. Additionally, the resting heart rate (HRmin) of the participants was measured, while the maximal heart rate was determined using the well-known formula: HRmax = 220 – age.

2.2.1 Pre-fatigue protocol

Participants completed two sets of five 1-min treadmill runs under varying conditions to analyze the effects of speed and incline on calcaneus kinematics. The first set (pre-fatigue) included trials at the following conditions: 10 km/h with no incline, 12 km/h with no incline, 14 km/h with no incline, 10 km/h with a 5° incline, and 10 km/h with a 10° incline. Each trial was followed by a 1-min rest (Figure 1). The order of trials was randomized to minimize any order effects. Participants were instructed to set the treadmill to the appropriate speed and incline before each run. After each run, there was a 1-min break.

2.2.2 Fatigue protocol

Following the pre-fatigue test protocol, participants underwent a fatigue phase that consisted of a 30-min run on a flat (no incline) at an intensity corresponding 80% of their heart rate reserve (HRR) (Figure 1). The target heart rate was calculated prior to the start of the measurement using the formula: HRR = [(HRmax–HRmin) × 0.8] + HRmin, where HRmin was resting heart rate. During the fatigue phase, participants self-regulated their running speed to maintain this target heart rate, aided by a heart rate monitor connected to the treadmill. After the fatigue phase, participants self-reported their fatigue level using a visual analog scale (VAS), where 0 indicated no fatigue and 10 indicated extreme fatigue. Thirteen out of fifteen participants reported a fatigue score of 6.4 ± 0.7, corresponding to a threshold between moderate and severe fatigue.

2.2.3 Post-fatigue protocol

After the fatigue run, participants repeated the same five 1-min trials (post-fatigue) under identical conditions as the pre-fatigue test protocol (Figure 1). The aim was to compare biomechanical differences in calcaneus kinematics before and after fatigue across the five test conditions.

Heart rate was measured only during the fatigue protocol, as the primary focus of the study was on ankle kinematics during the 5-run protocol. The short duration of the runs minimized any potential impact of cardiovascular stress on the kinematic results. Furthermore, VAS scores following the repeated 5-run protocol (6.1 ± 1.1) indicate that the short runs did not significantly affect participant fatigue. This is supported by the lack of a statistically significant difference between VAS scores after the fatigue protocol and after the 5-run protocol ($p = 0.34$).

2.3 Monitoring

2.3.1 Heel strike foot pattern

A DS-CAM-1100m high-frequency camera set at 300fps, an 8-channel 24 bit and 200 kHz Dewe 43 DAQa with DewesoftX software environment (all DEWESoft, Slovenia) were used to analyse running technique, excluding subjects who did not run over the heel.

2.3.2 Running

The running protocol was performed on a TRX Marathon treadmill (Toorx, Pozzolo Formigaro, Italy) with a belt size of 530 × 1520 mm. The treadmill allowed adjustable speeds from 0.8 to 22.0 km/h and an incline from 0% to 13%. The kinematic data acquisition was performed with a Qualisys Oqus system which consists of 12 infrared cameras (Qualisys AB, Gothenburg, Sweden) operating at a recording frequency of 180 Hz. Reflective markers, each with a diameter of 14 mm (Qualisys AB, Gothenburg, Sweden), were used for the measurements. A total of 43 reflective markers were attached to the body prior to the measurement. Two marker sets were utilized: the Qualisys Sports Marker Set and the IORfoot Marker Set for the foot and lower limbs. From the Qualysis Marker Set we used all the markers while from the IORfoot Marker Set we used only 4 markers that are marked with red circles in Figure 2A. In addition, 4 passive markers were attached to a treadmill in order to calibrate the treadmill in a laboratory. The heart rate was monitored with a Polar H10 device (Polar Electro Oy, Kempele, Finland).

2.4 Running shoes

To obtain accurate measurements of the calcaneus angle, the markers must be placed directly on the skin and not on the

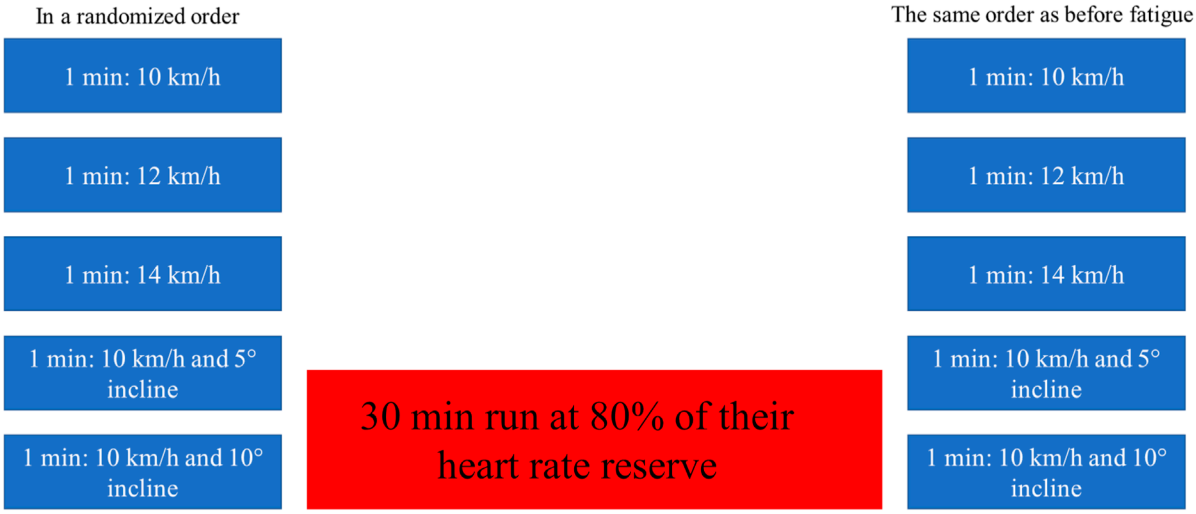

FIGURE 1
Experimental protocol.

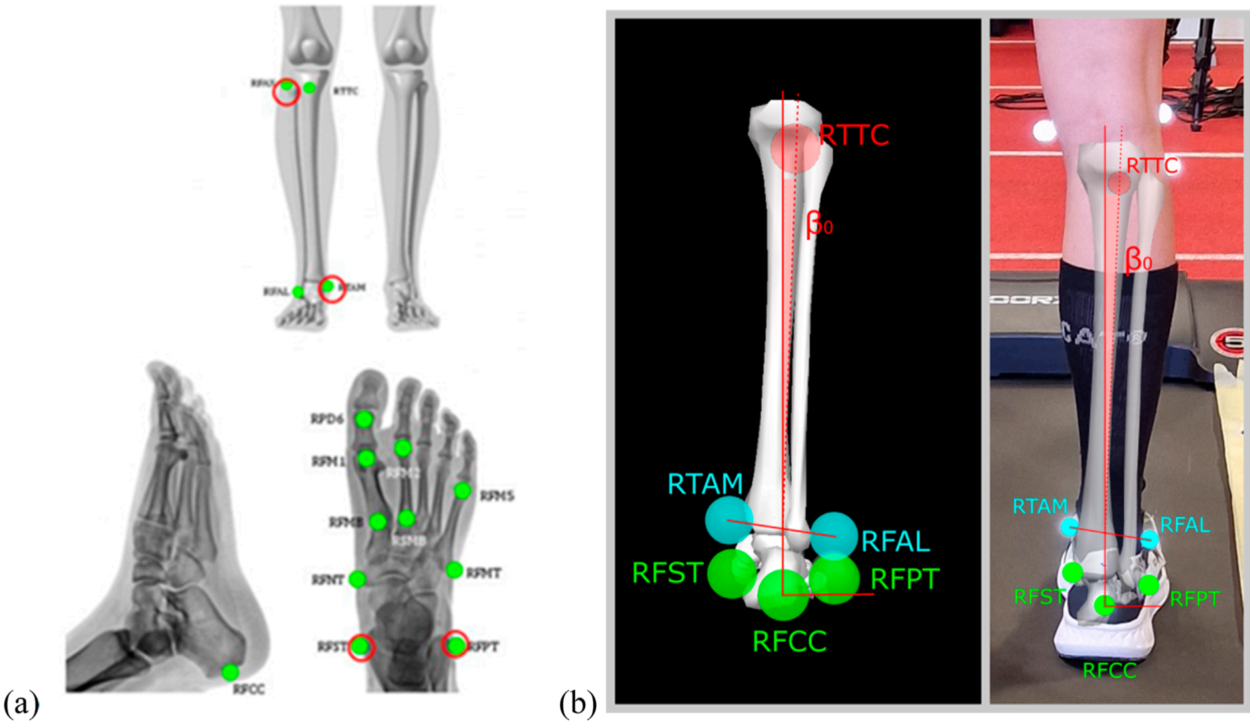

FIGURE 2
(A) Placement of "IORFoot Model" passive markers, **(B)** passive markers and presentation of eversion angle.

shoe. Therefore, our running shoes have been specially modified to make this possible. In collaboration with Alpina (Alpina, Žiri, Slovenia), the shoes were redesigned to have small openings in precise locations—under the medial malleolus and around the calcaneus area—to allow the passive reflective markers to be placed directly on the runner's skin, Figure 3. These strategic openings ensured that the movement of the calcaneus could be accurately measured while still enabling good fixation and comfort of the running shoes. This was a crucial factor in the methodology of the present study to observe the natural calcaneus movement without the interference of the shoe material. Additionally, corresponding openings were created in the socks to ensure that the markers were securely attached directly to the skin.

FIGURE 3
Adapted running shoes with modifications for attaching markers, with openings under the medial malleoli and the heel.

2.5 Data analysis

The kinematic data on the movement of the calcaneus were recorded using the Qualisys kinematic system. The associated Track Manager software enabled the creation of a model for the automatic detection and data acquisition of passive markers. This model enabled the recognition of marker positions in space and the subsequent creation of movement models for body segments. The orientation of the rigid body segments in space was determined by calculating the Euler angles using Visual 3D software (C-Motion, Maryland, United States). Rotations around the anterior-posterior axis (x-axis), specifically the calcaneus inversion/eversion angle, were crucial for the subsequent analysis. Following the guidelines of the International Society of Biomechanics (ISB), (Leardini et al., 2021), a static measurement was performed to establish a neutral position in which the eversion angle was set to 0° for each subject.

The data were then processed in the MATLAB R2021b (Mathworks, Massachusetts, United States) where algorithms recognised and differentiated steps based on the height of the heel markers and calcaneus angles calculated from Visual3D software. Foot contact with the ground was identified by analyzing the local maxima of the inversion angle prior to the minimum z-position of the heel marker, which also needed to be below a threshold value of 3.5 cm. The toe-off event was defined as a 3.5 cm ascent above the minimum z-position of the marker placed on the second toe. During ground contact, we assessed the eversion/inversion angle of calcaneus at heel strike (β_0), maximum eversion angle of calcaneus (β_{MAX}), and calcaneus range of motion–difference between subsequent parameters (β_{ROM}) for subsequent statistical analysis. Eversion angle is presented on Figure 2B.

Steps included in the analysis were selected from the end of the 1-min interval. To assess the reliability of the eversion angle measurements, we determined that an analysis of the final 10 s of the running trials, which typically corresponds to around 15 steps, provides a reliable estimate of the eversion angle with a SEM of 0.15°. This duration was chosen because by this point, participants had adapted to the running speed, minimizing variability from initial adjustments.

2.6 Statistical analysis

Statistical analyses were conducted using RStudio (Massachusetts, United States). Mixed models were employed to analyze the relationship between the selected response variables and the predictor variables. High inter-subject variability was observed in the sample, as indicated by the residual plots, which revealed heteroscedasticity. To address this, a random effect for individual subjects was included in the models. Initially, we assessed the assumptions of homoscedasticity and normality by examining the residual plots. The assumption of homoscedasticity was violated for all variables. This issue was resolved by incorporating a random subject effect at the intercept in the mixed models, which successfully addressed the violation and ensured compliance with the assumption of homoscedasticity.

Additionaly, standard error of the measurement (SEM) was calculated for all three parameters, i.e., eversion/inversion angle at heel strike, maximal eversion angle, and range of motion. Furthermore, we applied random noise of 1 mm uniformly across all three spatial coordinates to the original marker position data and subsequently reprocessed the data using the Visual3D pipeline to evaluate the effect of the 3D kinematic system error on the derived parameters.

3 Results

3.1 Influence of different speeds

Before fatigue (Figure 4; Table 1), variable speed had no statistically significant overall effect on β_0 $p = 0.07$). However, β_0 was significantly higher during running at 14 km/h compared to running at 10 km/h (1.8° ± 3.9° vs. 2.1° ± 3.4°, $p = 0.03$). There was no significant difference in β_0 between running at 12 km/h and 10 km/h ($p = 0.65$). In contrast, variable speed after fatigue (Table 2) had a statistically significant overall effect on β_0 ($p = 0.05$). In addition, β_0 was significantly higher when running at 14 km/h (3.4° ± 3.0° vs. 2.4° ± 3.4°, $p = 0.17$), but not when running at 12 km/h compared to 10 km/h (4.2° ± 4.6° vs. 2.4° ± 3.4°, $p = 0.01$).

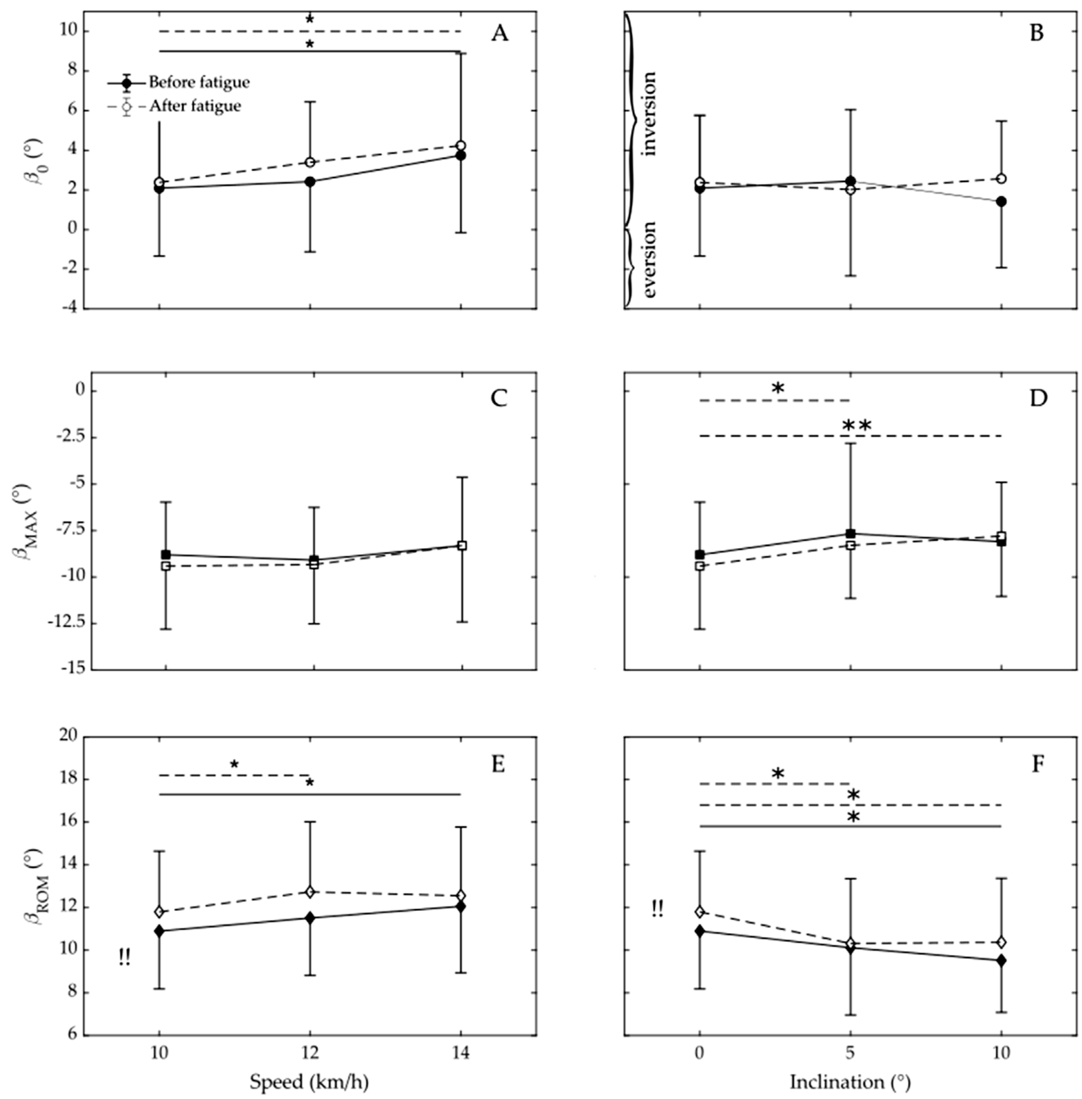

FIGURE 4
Calcaneus eversion/inversion angle at heel strike (β_0) across varying speeds **(A)** and inclinations **(B)**; maximum eversion angle ((β_{MAX}) at different speeds **(C)** and inclinations **(D)**; and range of motion in the calcaneus (β_{ROM}) for varying speeds **(E)** and inclinations **(F)**. Data prior to the onset of fatigue is indicated by filled circles (squares, diamonds) and a solid line, whereas data subsequent to fatigue is represented by open circles (squares, diamonds) and a dashed line.*Indicates significant difference (p < 0.05),**Indicates highly significant difference (p < 0.001), !! Indicates that fatigue has an overall significant effect on β_{ROM} (p < 0.05).

Variable speed had no statistically significant overall effect on β_{MAX} before (p = 0.47, Table 1)or after fatigue (p = 0.25, Table 2). There was also no statistically significant difference in β_{MAX} when running at 12 or 14 km/h compared to 10 km/h, either before (p = 0.66 and p = 0.44, respectively) or after fatigue (p = 0.91 and p = 0.13, respectively).

Variable speed had a statistically significant overall effect on β_{ROM} before (p = 0.02, Table 1) but not after fatigue (p = 0.09, Table 2). In addition, β_{ROM} was statistically significant when running at 14 km/h before fatigue and when running at 12 km/h after fatigue compared to 10 km/h (10.9° ± 2.7° vs. 12.1 ± 3.1, p = 0.004, and 11.8 ± 2.8 vs. 12.7 ± 3.3*, p = 0.04, respectively).

3.2 Influence of different inclinations

Before and after fatigue (Figure 4; Tables 1, 2), the variable inclination had no statistically significant overall effect on β_0 (p = 0.57 and p = 0.52, respectively). Before fatigue (Table 1), β_0 during running at 5° or 10° also did not differ significantly from that with no incline (p = 0.68 and 0.08, respectively). Even after fatigue (Table 2), β_0 during running with an inclination of 5° or 10° did not differ significantly from that with no incline (p = 0.46 and 0.70, respectively).

Before fatigue (Table 1), the inclination had no overall statistically significant effect on β_{MAX} (p = 0.42). After fatigue (Table 2), however, the inclination had a statistically

TABLE 1 Three-dimensional kinematics of the lower leg during running prior to the onset of fatigue. Presented are the group means ± standard deviation (SD).

Before fatigue	10 km/h	12 km/h	14 km/h	10 km/h, 5° incline	10 km/h, 10° incline
β_0 [°]	2.1 ± 3.4	2.4 ± 3.5	3.8 ± 3.9*	2.4 ± 4.8	1.4 ± 3.3
β_{MAX} [°]	−8.8 ± 2.8	−9.1 ± 2.8	−8.3 ± 3.7	−7.7 ± 4.9	−8.1 ± 3.2
β_{ROM} [°]	10.9 ± 2.7	11.5 ± 2.7	12.1 ± 3.1*	10.1 ± 3.2	9.5 ± 2.4†

*Statistically significant compared to 10 km/h ($p < 0.05$), †Statistically significant compared to 0° inclination ($p < 0.05$).

TABLE 2 Three-dimensional kinematic analysis of the lower leg during running post-fatigue. The results are presented as group means ± standard deviation (SD).

After fatigue	10 km/h	12 km/h	14 km/h	10 km/h, 5° incline	10 km/h, 10° incline
β_0 [°]	2.4 ± 3.4	3.4 ± 3.0	4.2 ± 4.6*	2.1 ± 4.0	2.6 ± 2.9
β_{MAX} [°]	−9.4 ± 3.4	−9.3 ± 3.2	−8.3 ± 4.1	−8.3 ± 2.9†	−7.8 ± 3.2††
β_{ROM} [°]	11.8 ± 2.8	12.7 ± 3.3*	12.6 ± 3.2	10.3 ± 3.0†	10.4 ± 3.0†

*Statistically significant compared to 10 km/h ($p < 0.05$), †Statistically significant compared to 0° inclination ($p < 0.05$), †Statistically significant compared to 0° inclination ($p < 0.001$).

significant effect on β_{MAX} (p = 0.002). In addition, β_{MAX} was significantly lower when running at 5° or 10° than with no incline (−9.4° ± 3.4° vs. −8.3° ± 2.9°, p = 0.01, and −9.4° ± 3.4° vs. −7.8° ± 3.2°, p = 0.0003).

Before and after fatigue (Tables 1, 2), inclination had an overall statistically significant effect on β_{ROM} (p = 0.02, and 0.003, respectively). In addition, β_{ROM} was significantly lower when running with a 10° incline before or after fatigue compared to no incline (9.5° ± 2.4° vs. 10.9° ± 2.7°, p = 0.004 and 10.4° ± 3.0° vs. 11.8° ± 2.8°, p = 0.003, respectively). Moreover β_{ROM} after fatigue was also significantly different when running with a 5° incline compared to running with no incline (10.3° ± 3.0° vs. 11.8° ± 2.8°, p = 0.002).

3.3 Influence of fatigue

Overall, fatigue (Table 2) had no significant effect on β_0 (p = 0.70). Furthermore, fatigue did not have a significantly different effect on β_0 at speed of 12 km/h or 14 km/h compared to 10 km/h (p = 0.52 and p = 0.85, respectively). Moreover, fatigue did not have a significantly different effect on β_0 at incline of 5° or 10° compared to no incline (p = 0.50 and p = 0.41, respectively).

Fatigue (Table 2) had no significant overall effect on β_{MAX} (p = 0.38). Furthermore, fatigue did not have a significantly different effect on β_{MAX} at speed of 12 km/h or 14 km/h compared to 10 km/h (p = 0.71 and p = 0.33, respectively). Moreover, fatigue did not have a significantly different effect on β_{MAX} at incline of 5° or 10° compared to no incline (p = 0.54 and p = 0.99, respectively).

Fatigue (Table 2) had an overall significant effect on β_{ROM} (p = 0.05). However, fatigue did not have a significantly different effect on β_{ROM} at speed of 12 km/h or 14 km/h compared to 10 km/h (p = 0.59 and p = 0.50, respectively). Moreover, fatigue did not have a significantly different effect on β_{MAX} at incline of 5° or 10° compared to no incline (p = 0.27 and p = 0.94, respectively).

3.4 Reliability and accuracy of the measurements

Standard error of measurement (SEM) for β_0, β_{MAX} and β_{ROM} were calculated to be 0.08°, 0.06°, and 0.05° respectively. Furthermore, we assessed the effect of random noise introduced into the data, and our findings showed that the SEM increased slightly, with values of 0.09°, 0.06°, and 0.06°. Additionaly, the variability of these angles was 0.7°, 0.6°, and 0.5°.

4 Discussion

The main aims of the present study were to determine whether different speeds, inclinations, or fatigue affected the calcaneus eversion/inversion angle at heel strike, the maximum eversion angle, or the range of motion of the calcaneus inversion/eversion angle. Findings in the present study indicate that speed significantly influenced the eversion/inversion angle at heel strike when participants were fatigued. Inclination had a statistically significant effect on maximum eversion angle after fatigue and on range of motion both before and after fatigue. Additionally, fatigue itself had a significant impact on β_{ROM}.

The reliability of the measured angles is indicated by the low Standard Error of the Mean (SEM) values, which are below 0.1°. The accuracy of the measurements was assessed by calculating the SEM while applying a uniform random noise of 1 mm across all three spatial coordinates to the original marker position data. The data were then reprocessed using the Visual3D pipeline to evaluate the impact of 3D kinematic system error on the derived parameters. The SEM exhibited a slight increase compared to the SEM calculated without the additional noise. However, this increase was minimal, indicating that the measurement error associated with the introduced noise did not significantly compromise the accuracy of the measurements. Therefore, we can conclude that the measurement system maintains a high level of accuracy, even in the presence of noise. All previously mentioned parameters are important regarding the prevention of RRI. As noted in prior publications $ß_0$ has been linked to RRIs, including iliotibial band syndrome (Grau et al., 2011) and Achilles tendinopathy. (McCRORY et al., 1999; Mousavi et al., 2019). However, existing literature comparing injured and non-injured runner groups presents conflicting results. Some studies indicate that uninjured runners exhibit a greater $ß_0$ (Kuhman et al., 2016; Hreljac et al., 2000), while other studies suggest the opposite (Grau et al., 2008) or report no significant difference (Hamacher et al., 2016). Additionally, $ß_{MAX}$ can increase strain on the Achilles tendon and other related structures, potentially leading to conditions such as medial tibial stress syndrome. (Stacoff et al., 2000; Ye et al., 2024). Willems et al. (2006) showed that an increased $ß_{MAX}$ raises the risk of exercise-related lower leg pain. In their study, $ß_{MAX}$ was significantly higher in injured participants (9.6° ± 5.9°) compared to uninjured participants (7.7° ± 5.1°), with the difference being statistically significant ($p = 0.03$). Furthermore, $ß_{ROM}$ is correlated with chronic ankle instability (CAI) (Cordova et al., 2000; Stotz et al., 2021). Individuals with a higher $ß_{ROM}$ may therefore be at an increased risk, suggesting that a greater $ß_{ROM}$ could predict a higher likelihood of RRI.

In the current study, the pre-fatigue and after-fatigue values of $ß_0$ (ranging from 2.1° to 3.4° and 2.4°–4.2°) are consistent with values reported in previous research: 2.7° for the left leg and 2.5° for the right leg by Goetze (2015), 4° by Hamacher et al. (2016), 3.5° by Stacoff et al. (2000) In all of the mentioned studies, markers were placed directly on the shoes rather than on the skin; however, as a previous study demonstrated (Sinclair et al., 2013) this should not have affect the values of $ß_0$, although it does have a significant impact on $ß_{MAX}$ and consequently on $ß_{ROM}$. As mentioned above Willems et al. (2006) measured $ß_{MAX}$ in injured participants at 9.6° ± 5.9° and 7.7° ± 5.1° in unijured participants. In the current study, $ß_{MAX}$ in uninjured participants ranged between 8.3° and 9.4° when running on a flat surface The slight discrepancy in these values could be attributed to the participants in the present study running in footwear, while those in the previously mentioned study ran barefoot; this is in agreement with a previous study by Thompson et al. (2015). This finding is consistent with the results of Stacoff et al. (2000) who used intracortical Hofmann pins to measure $ß_{MAX}$ in barefoot and shod running, $ß_{MAX}$ of 6.9° ± 0.7° and 8.8° ± 1.5°, respectively. Therefore, we can confirm that values in the present study align with the literature for uninjured runners. The $ß_{ROM}$ observed in the present study is comparable to that reported by Hein and Grau, (Hein and Grau, 2014), who measured $ß_{ROM}$ while participants ran either in shoes or barefoot at a speed of 11 km/h. They found a $ß_{ROM}$ of 12° ± 3° when running in shoes, which is similar to the 10.9° ± 2.7° and 11.5° ± 2.7° $ß_{ROM}$ we observed at speeds of 10 and 12 km/h, respectively.

4.1 Influence of different speeds on $ß_0$, $ß_{MAX}$, and $ß_{ROM}$

In the current study, running at 14 km/h led to a significantly higher $ß_0$ compared to running at 10 km/h, both prior to and following fatigue. A higher $ß_0$ could be attributed to a greater proportion of foot placement on the middle and front foot as running speed increases (Keller et al., 1996). This is consitetn with the results published by Koldenhoven et al. (2019) observed across 3 walking speeds—preferred walking speed, 120% of preferred walking speed, and a standardized faster speed. Their findings indicated that participants with chronic ankle instability displayed greater inversion at higher speeds, while those who had previously experienced an ankle sprain but had returned to pre-injury function demonstrated increased eversion at greater speeds.

Contrary, there was no statistically significant difference in $ß_{MAX}$ when participants ran at different speeds. Although there was a noticeable trend indicating a smaller $ß_{MAX}$ when running at 14 km/h compared to 10 km/h, both in fatigued and non-fatigued states, this trend was not statistically significant. This finding is consistent with the study by Koldenhoven et al. (2019), Conversely, another study indicated an increased $ß_{MAX}$ with increased speed; (Fukuchi et al., 2017); however, that study placed markers directly on the running shoes, thus measuring the eversion of the foot rather than that of the calcaneus. As mentioned above, $ß_{MAX}$ is significantly different when markers are placed on the shoes compared to when they are placed directly on the skin (Sinclair et al., 2013).

Moreover, speed had a significant overall effect on $ß_{ROM}$ in the non-fatigued state, with higher speed leading to an increased range of motion. This finding aligns with the results from Fukuchi et al. (2017), who examined running at three different speeds (9, 12.6, and 16.2 km/h) and observed that higher speeds were associated with an increased range of motion. However, in the fatigued state in our study, only running at 12 km/h resulted in a significant increase in $ß_{ROM}$ compared to running at 10 km/h.

As previously discussed, an increased $ß_{ROM}$ is associated with CAI, which is a known predictor of RRI. Furthermore, although we observed a trend toward a higher $ß_{MAX}$ with increased running speed, this trend did not reach statistical significance. Nevertheless, these findings suggest that higher running speeds may elevate the risk of RRIs, warranting caution and further investigation.

4.2 Influence of different inclinations on $ß_0$, $ß_{MAX}$, and $ß_{ROM}$

In the present research, we found that inclination did not significantly affect $ß_0$, which aligns with the findings of Sinclair et al. (2018) Additionally, $ß_0$ before fatigue we observed at inclinations of 0°, 5°, and 10° are comparable to those reported in the previous study (2.9° vs. 2.1°, 3.7° vs. 2.4°, and 2.2° vs. 1.4°, respectively) (Sinclair et al., 2018). However, Dixon et al. (2011) reported a higher $ß_0$ when running uphill at a 10° incline. The key difference between Dixon et al.'s study and the studies conducted by Sinclair and the present study is that the participants in Dixon's study were running barefoot, whereas in both Sinclair's study and the current, the participants were running in shoes. Therefore, we can conclude that uphill running in running shoes does not significantly alter the inversion angle at foot strike.

In the non-fatigued state, inclination had no statistically significant effect on $ß_{MAX}$. However, when participants were fatigued, $ß_{MAX}$ was lower when running uphill at either 5° or 10°. This suggests that running uphill in a fatigued state may reduce the risk of injury due to excessive calcaneal eversion. This finding for non-fatigued state is consistent with that of Sinclair et al. (2018), who found no statistically significant difference in $ß_{MAX}$ at different inclinations.

Additionally, we can speculate that the development of RRI may be reduced due to smaller $ß_{ROM}$ when running at an incline of 10° in both fatigued and non-fatigued states, and at an incline of 5° when in a fatigued state, where the results were statistically significant. However, Sinclair et al. (2018) found no statistically significant difference in this parameter when evaluating different inclines. There was an observable trend, with $ß_{ROM}$ recorded at 11.21° ± 5.59° for a 5° incline and 9.89° ± 4.16° for a 15° incline; however, as previously noted, this difference was not statistically significant. The absolute values of $ß_{ROM}$ were comaprable in both studies (11.21° ± 5.59° vs. 10.1 ± 3.2 at the 5° incline).

4.3 Influence of fatigue on $ß_0$, $ß_{MAX}$, and $ß_{ROM}$

Fatigue did not affect the $ß_0$ across different speeds or inclinations in present study. These observations are consistent with a previous study, (Van Gheluwe and Madsen, 1997), which also found that $ß_0$ remained unchanged before and after fatigue. In that study, markers were placed on the shoes, which, as mentioned earlier, differs from placing markers directly on the skin (Sinclair et al., 2013).

In addition, fatigue did not have a significant effect on $ß_{MAX}$ across different speeds or inclinations. Although there was a visible trend toward higher $ß_{MAX}$ after fatigue when running at 10 or 12 km/h and at a 5° incline, the differences were not statistically significant. We initially expected that $ß_{MAX}$ would change after fatigue, as previous studies have suggested that foot and joint kinematics are altered at key points in the running cycle due to fatigue (Van Gheluwe and Madsen, 1997; Elliott and Roberts, 1980). To our knowledge, only one study has measured $ß_{MAX}$ during shod running in fatigued state with markers attached directly to the skin (Brown et al., 2014). However, this study did not examine the effect of fatigue on $ß_{MAX}$ but rather compared the dominant and non-dominant limbs before and after fatigue. As a result, it remains unclear whether fatigue has a statistically significant impact on $ß_{MAX}$. Nonetheless, their results are similar to present study, showing a change in eversion angle of 1.8° in the dominant limb and 0.6° in the non-dominant limb after fatigue during running at 12 km/h. In present study, the change was slightly smaller, at just 0.2°. Another study (Van Gheluwe and Madsen, 1997) showed that measured $ß_{MAX}$ found a statistically significant difference after fatigue, but the markers in that study were placed on the shoes, which does not directly reflect calcaneus motion, as mentioned earlier. It is also possible that participants in the present study, who were required to be active for at least 5 h per week, were well-trained, and the 0.5-h exhaustion run may not have been sufficient to induce significant fatigue.

Fatigue has a substantial impact on $ß_{ROM}$, indicating that running while fatigued may elevate the risk of developing RRI. This finding aligns with the research conducted by Koblbauer et al. (2014) which demonstrated that the range of motion in the non-dominant leg was significantly higher in a fatigued state compared to pre-fatigued conditions. While the running speed was not reported in their study, the results are consistent with our findings for running at 12 km/h (11.5° ± 3.9° vs. 11.5° ± 2.7° in the non-fatigued and 13.1° ± 4.6° vs. 12.7° ± 3.3° in the fatigued state).

5 Conclusion

This study aimed to explore the impact of different running speeds, inclines, and levels of fatigue on the calcaneus eversion/inversion angle at heel strike, maximum eversion angle and range of motion during running in injury-free female runners. The findings indicate that speed significantly influences $ß_0$, particularly when participants are fatigued. Inclination also had a statistically significant effect on $ß_{MAX}$ after fatigue and $ß_{ROM}$ both before and after fatigue. Additionally, fatigue itself was found to significantly impact the calcaneus $ß_{ROM}$, The findings suggest that running at higher speeds and in a fatigued state increases the likelihood of RRI due to the higher $ß_{ROM}$, while running at an incline may reduce this risk by lowering the $ß_{ROM}$ as well as $ß_{MAX}$. These results underscore the importance of considering speed, incline, and fatigue in injury prevention strategies for runners with heel-to-toe foot strike. However, it is important to note that these conclusions are specific to how the kinematics of the calcaneus can influence the risk of RRI.

6 Limitations

The findings of the study should be interpreted with some caution due to its limitations. Firstly, the sample size was relatively small, comprising only 15 female participants. This restricts the generalizability of the results to a wider population, including male runners. The decision to focus on female runners was based on observed differences in injury prevalence and biomechanics between genders, particularly in lower limb injuries related to running (Rubio et al., 2023; Xie et al., 2022). However, future studies should consider including male participants to determine whether the findings are consistent across genders. Secondly, this study focused only on injury-free participants, which may not fully capture the variability in running biomechanics that is seen in individuals with a history of running-related injuries (RRIs). It is well known that individuals with previous injuries often demonstrate altered running patterns and altered kinematics, which could impact the results. While the inclusion of injury-free participants helps standardize the biomechanical measurements, the results may not fully reflect the injury risk for those with past injuries. Additionaly, the study exclusively included runners with a heel-to-toe foot strike pattern. While this is common among distance runners, it does not account for the biomechanics of runners who adopt a midfoot or forefoot strike. Different foot strike patterns could result in different kinematic patterns, and thus, future studies should examine whether the observed findings are consistent across a variety of foot strike patterns. Furthemore, the fatigue protocol involved a 30-min run, which may not have resulted in the same level of fatigue as longer or more intense running sessions. Further research with a larger and more diverse sample, as well as varying fatigue protocols, is necessary to validate and expand upon these findings. The study was conducted in a controlled laboratory setting, which may not fully replicate the conditions encountered by runners in real-world environment. Finally, the use of reflective markers to capture kinematic data can also present limitations, despite efforts to minimize errors by affixing markers directly to the skin within the customized running shoes. Although this approach reduces errors commonly associated with marker placement on shoes, there can still be inaccuracies in attaching the markers to anatomical landmarks despite a low standard error was observed in the studied parameters.

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 humans were approved by the study was approved by the Committee for Ethical Issues in Sports at the University of Ljubljana, Slovenia (1/2023). The studies were conducted in accordance with the local legislation and institutional requirements. The participants provided their written informed consent to participate in this study.

Author contributions

NV: Conceptualization, Formal Analysis, Funding acquisition, Visualization, Writing – original draft, Writing – review and editing. NN: Data curation, Formal Analysis, Methodology, Writing – review and editing. MD: Conceptualization, Data curation, Formal Analysis, Funding acquisition, Writing – review and editing. IP: Data curation, Formal Analysis, Writing – review and editing. VS: Conceptualization, Funding acquisition, Writing – review and editing. MS: Conceptualization, Funding acquisition, Writing – review and editing.

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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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Patellar cartilage thickness relates to knee external rotation during squatting in individuals with and without patellofemoral pain—a pilot study

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The relationship between patellofemoral cartilage morphology and knee external rotation (KER), one of the possible factors increasing patellar cartilage stress, has been rarely explored in individuals with and without patellofemoral pain (PFP). Ten individuals with PFP and 10 pain-free controls, matched for age, weight, height, and activity level, participated. Patellar cartilage morphology was assessed using 3-Tesla magnetic resonance imaging. Lower extremity kinematics during bilateral squatting at 45° of knee flexion were captured using a 3-dimensional motion capture system. Pearson and Spearman correlation coefficients were used to assess the associations between cartilage thickness (medial, lateral, and total) and peak KER, along with other peak joint angles across the three planes. Across all participants, there were significantly moderate correlations between medial cartilage thickness and KER ($r = -0.48$, $p = 0.03$), and total cartilage thickness and KER ($r = -0.47$, $p = 0.35$). In the PFP group, there was a significantly large correlation between medial cartilage thickness and KER ($r = -0.66$, $p = 0.03$). In the control group, there was a significant very large correlation between lateral cartilage thickness and KER ($r = -0.79$, $p = 0.01$) and a significant very large correlation between total cartilage thickness and KER ($r = -0.75$, $p = 0.01$). The findings suggest that thinner patellar cartilage is associated with increased KER during bilateral squatting in persons with and without PFP. Since our study focused on a double-limb activity, which may require less KER, future research should examine its impact on cartilage morphology during single-limb activities.

KEYWORDS

tibial rotation, patellofemoral pain, anterior knee pain, patellar cartilage, cartilage morphology, knee rotation

1 Introduction

Patellofemoral pain (PFP) refers to pain localized around the peripatellar and/or retropatellar regions, commonly occurring during activities that involve knee flexion under weight bearing, such as running and squatting (1, 2). The causes of PFP include irritation of the innervated structures of the patellofemoral joint (PFJ), such as the subchondral bone, often due to increased pressure on the cartilage of the PFJ (3–5). Repeated excessive loading of the PFJ during weight-bearing activities can lead to cartilage thinning and altered composition (6). Farrokhi et al. (6) have shown that individuals with PFP exhibit thinner patellar cartilage compared to pain-free controls.

Thinner PFJ cartilage can increase stress on the cartilage (7), as it reduces the cartilage's ability to absorb and distribute force during weight-bearing. Additionally, a computational modeling study found a significant negative association between patellar bone strain and cartilage thickness, suggesting that thinner patellar cartilage may contribute to bone injuries and pain in individuals with PFP (8).

It is important to note that faulty lower extremity kinematics can contribute to increased stress on the PFJ. Specifically, in addition to hip adduction and internal rotation (9), knee flexion (10) and abduction (9), knee external rotation (KER) is reported as a contributing factor leading to elevated PFJ stress in a cadaveric study (11). It is believed that external rotation of the tibia shifts the patellar contact region laterally and increases the contact pressure on the lateral aspect of the patella (12). Specifically, KER during activities causes the tibial tuberosity to move laterally, which in turn pulls the patella laterally (13). This movement results in a malalignment between the patella and the trochlear groove of the femur, leading to improper patellar tracking (13). Consequently, the contact pressure on the lateral facet of the patella increases, with force being concentrated on the lateral aspect of the patella, thus elevating the stress in this region (13, 14). A finite element analysis demonstrated that a 10° increase in KER elevates average patellar cartilage stress by 11% in pain-free females (15). Additionally, KER has been identified as a key predictor of patellar cartilage stress during running in both pain-free individuals and those with PFP (16). Collectively, findings from these studies (11–16) suggest that KER during movements may increase PFJ stress, potentially contributing to a reduction in patellar cartilage thickness.

A key limitation of previous studies is that their findings were based on cadaveric specimens or computational models that simulate human movements and PFJ geometry (11–13). As a result, the relationship between cartilage morphology and KER in living participants remains unclear. Furthermore, while some modeling studies have incorporated subject-specific movements and cartilage geometry (15, 16), they did not evaluate whether the medial or lateral cartilage was more affected. This gap makes it uncertain which regional cartilage morphology is more strongly associated with KER. Lastly, the limited number of studies involving human participants further complicates our understanding of how cartilage morphology relates to KER during tasks that commonly exacerbate PFP, such as squatting, particularly across diverse populations.

To date, no studies have been conducted to examine how KER during weight-bearing activities relates to patellofemoral cartilage in persons with PFP, despite previous studies suggesting a potential link between KER and cartilage morphology in this population. Thus, the aim of this study was to investigate the association between patellar cartilage thickness and lower extremity kinematics, particularly KER during bilateral squatting, in individuals with and without PFP. Bilateral squatting was chosen for this study due to its functional relevance and better tolerance in individuals with PFP (1, 2). We hypothesized that thinner patellar cartilage would be associated with increased KER during bilateral squatting in individuals with and without PFP, suggesting a potential link between patellar cartilage morphology and knee frontal plane kinematics.

2 Methods

2.1 Participants

This is a cross-sectional study involving a post-analysis of data from 20 young female participants (10 PFP and 10 pain-free controls) previously enrolled in an earlier study (8). There were no significant differences in age, height, weight, and activity level between groups (Table 1). Participants in the PFP group were eligible if they experienced a gradual onset of retropatellar pain for at least 3 months (8). Informed consent was obtained from all participants, in accordance with the guidelines of the Health Sciences Institutional Review Board at the University of Southern California.

For participants in the PFP group, a physical examination was conducted to ensure other sources of pain were excluded (8). This evaluation included palpating the soft tissues surrounding the PFJ to accurately localize the pain. Participants with PFP were excluded if their pain originated from areas such as the quadriceps tendon, patellar tendon, patellar bursa, fat pad, menisci, or tibiofemoral joint (8). Additional exclusion criteria for participants in the PFP group included prior knee surgery, history of traumatic patellar dislocation, or the presence of implanted devices that could interfere with the magnetic field of the magnetic resonance imaging (MRI) (8).

The control group was matched to the PFP group in terms of age, height, weight, and activity levels (within 10% difference). Physical activity was assessed using the World Health Organization's Global Physical Activity Questionnaire (17). Control participants met the same selection criteria as the PFP group, except they had no history of PFP.

2.2 Procedures

Data collection occurred in two phases: MRI and biomechanical testing. In the PFP group, MRI scans were

TABLE 1 Participants characteristics (mean ± standard deviation).

Characteristics	Patellofemoral pain group (n = 10)	Control group (n = 10)	P-value
Age (years)	25.1 ± 4.7	25.8 ± 6.1	0.78
Height (m)	1.77 ± 0.1	1.77 ± 0.1	0.97
Weight (kg)	59.7 ± 9.3	59.8 ± 7.1	0.45
Activity level (MET.min/week)	2,166.0 ± 969.1	1,944.0 ± 859.0	0.59

obtained from the affected limb; if both knees were symptomatic, the more painful side was selected. For control participants, the limb selected for MRI was matched to the corresponding side of their PFP counterpart. During biomechanical testing, data were collected from both limbs in all participants. However, for the PFP group, only data from the painful or more painful limb were analyzed. In the control group, analysis was conducted on the matched limb. This is to ensure consistency in assessing cartilage morphology and kinematics within the same limb for all participants.

2.2.1 MRI assessment

To obtain patellar cartilage thickness, all participants were seated in a wheelchair with their legs raised to relieve pressure from the knee joint for at least 60 min prior to imaging. This procedure was used to prevent any external compression that might influence cartilage deformation (6, 18). All participants with PFP and pain-free controls underwent an MRI scan using a 3.0 Tesla General Electric scanner (GE Healthcare, Milwaukee, WI, USA). The patellar cartilage was assessed using axial plane images of the PFJ, acquired with a 3-dimensional (3D) fast Spoiled Gradient Recalled Echo (SPGR) sequence (TR = 16.3 ms, TE = 2.8 ms, flip angle = 10°, matrix = 384 × 160, field of view = 160 × 160 mm, slice thickness = 2 mm, knee positioned at 0° flexion, total scan time = 2 min).

2.2.2 Biomechanical testing

During biomechanical testing, we aimed to quantify lower extremity kinematics during a bilateral squat. This task was selected because squatting is a common movement that often elicits pain in individuals with PFP (1, 2). Kinematic data were collected using an 11-camera Qualisys motion analysis system operating at 60 Hz (Qualisys Inc., Gothenburg, Sweden).

Before the testing, reflective markers were applied to all participants by the same investigator, a physical therapist. The anatomical markers were positioned on key bony landmarks, including the 1st and 5th metatarsal heads, medial and lateral malleoli, medial and lateral femoral epicondyles, the L5-S1 joint space, and bilaterally on the greater trochanters, and iliac crests, anterior superior iliac spines (ASISs). Rigid quadrilateral marker clusters were attached bilaterally on the lateral thigh and lower leg. Additionally, triads of rigid markers were placed on the heel counters of the shoes. After the markers were applied, a standing calibration trial was conducted to establish segmental coordinate systems and joint axes. Following this calibration, all anatomical markers were removed except for those at the iliac crests and the L5-S1 junction. The tracking clusters remained in place throughout the data collection session (19).

Following a standing calibration trial, participants were instructed to perform a bilateral squat, maintaining 45° of knee flexion. Participants were asked to maintain an upright posture for 10 s, holding their arms extended forward and keeping fingertip contact with a pole (8).

2.3 Data analysis

2.3.1 Cartilage thickness

The medial and lateral patellar cartilage was manually segmented by a trained investigator using commercial software (sliceOmatic, Tomovision, Montreal, Québec, Canada) (Figure 1). Cartilage thickness was calculated by measuring the perpendicular distance between opposing voxels that defined the boundaries of the patellar cartilage, using a custom MATLAB program (MathWorks, Natick, Massachusetts). The medial and lateral patellar cartilage thicknesses were determined as the average distance within each region of interest. Total cartilage thickness was calculated as the mean of the measurements from both regions (6).

We selected three regions of the patellar cartilage—medial, lateral, and total—due to their distinct anatomical and mechanical roles. Individuals with PFP often exhibit patellar cartilage thinning, particularly on the lateral facet (6, 8). Given our focus on the relationship between KER and patellar cartilage, assessing region-specific cartilage morphology is essential. Additionally, total cartilage thickness serves as a comprehensive indicator of overall patellar cartilage health.

2.3.2 Lower extremity kinematics

Reflective markers were labeled and digitized using Qualysis software (Qualisys Inc., Gothenburg, Sweden). Kinematic data for the hip, knee, and ankle in the sagittal, frontal, and transverse planes were quantified using Visual3D software (C-Motion, Rockville, MD). The kinematic data were filtered with a low-pass Butterworth filter (cutoff frequency: 6 Hz). Peak values of hip flexion, internal rotation, and adduction; knee external rotation

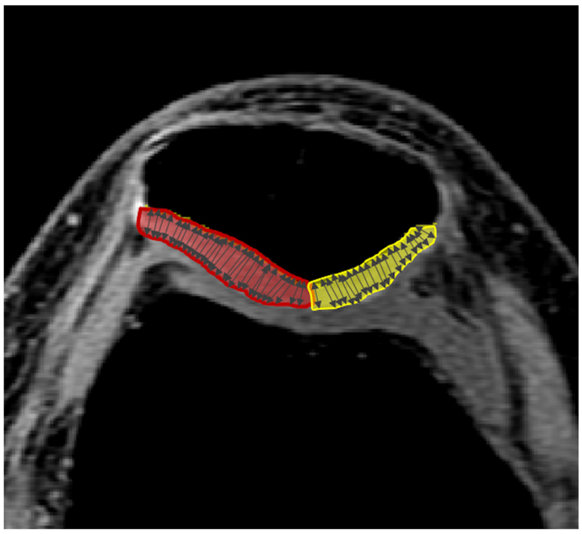

FIGURE 1
Quantification of patellar cartilage thickness. The red and yellow regions highlight the segmented lateral and medial patellar cartilage, respectively. Arrows indicate the measurements of the perpendicular distances between opposing voxels that define the boundaries of the patellar cartilage.

and abduction; and ankle dorsiflexion, pronation, and external rotation during squatting were extracted for analysis.

2.4 Statistical analysis

All statistical analyses were conducted using IBM SPSS Statistics (version 28; IBM Corp., Armonk, NY). The Shapiro–Wilk test was used to assess the normality of the kinematic data and cartilage thickness. Except for the peak ankle inversion angle, which was not normally distributed, all variables showed normal distributions. Therefore, a non-parametric Spearman's rank correlation was used to assess the relationship between cartilage thickness and peak ankle inversion, while Pearson's correlation coefficient was applied for all other correlations between cartilage thickness and kinematic variables. Correlations were analyzed separately for participants with PFP, pain-free participants, and the combined group. Correlation coefficients were categorized as follows: small ($0.1 < r \leq 0.3$), moderate ($0.3 < r \leq 0.5$), large ($0.5 < r \leq 0.7$), very large ($0.7 < r \leq 0.9$), and extremely large ($r > 0.9$) (20). *T*-tests or Mann–Whitney *U*-tests were used to compare kinematic variables and cartilage morphology between groups, depending on the normality of the data distribution. Statistical significance was set at $p < 0.05$ for all analyses.

3 Results

Statistically significant differences between the two groups were found only in cartilage thickness, while lower extremity kinematics showed no significant differences (Table 2). Across all participants, there was a significant moderate negative correlation between KER and both medial and total cartilage thickness (Figure 2). In individuals with PFP, the medial cartilage thickness showed a significantly large negative correlation with KER ($r = -0.66$ $p = 0.037$). In the control group, significant very large negative correlations existed between the total cartilage thickness and KER ($r = -0.75$ $p = 0.013$), and between the lateral cartilage thickness and KER ($r = -0.79$ $p = 0.007$). In addition, the medial and total cartilage thicknesses showed significant, very large ($\rho = -0.70$, $p = 0.025$) and large ($\rho = -0.67$, $p = 0.033$) negative correlations with ankle inversion, respectively.

No statistically significant correlations were observed between patellar cartilage thickness and hip movements in the sagittal, transverse, and frontal planes, as well as knee frontal plane movement and ankle movements in the sagittal and transverse planes, in the PFP, control, or combined groups ($p > 0.05$).

4 Discussion

4.1 Overall associations between patellar cartilage thickness and KER

To the authors' knowledge, this is the first study to explore the relationship between patellar cartilage thickness and KER in individuals with PFP. We aimed to investigate the relationship between the cartilage thickness of the PFJ and lower limb kinematics in individuals with and without PFP. Consistent with our hypothesis, the findings indicated that decreased medial and total cartilage thicknesses were significantly associated with greater KER across all participants. Specifically, in individuals with PFP, thinner medial cartilage was linked to increased KER, while in the control group, thinner lateral and total cartilage thicknesses showed significant associations with greater KER. These results suggest that regardless of the existence of PFP, thinner patellar cartilage thickness significantly correlates with larger KER, known as a factor increasing stress on the PFJ (9, 10, 12, 16). Additionally, previous studies (6, 8) have reported that patellar cartilage thickness is significantly thinner in persons with PFP compared to those without. One possible explanation for this patellar cartilage morphological change is the cumulative

TABLE 2 Comparisons of lower extremity kinematics and patellar cartilage thickness between groups.

Variables		Mean ± SD or Median (IQR)		*P*-value
		Patellofemoral pain group (*n* = 10)	Control group (*n* = 10)	
Lower extremity kinematics (deg)	Hip flexion	37.42 ± 10.56	36.46 ± 15.46	0.872[a]
	Hip internal rotation	−4.24 ± 4.26	−6.23 ± 2.17	0.204[a]
	Hip abduction	2.74 ± 3.68	3.43 ± 3.84	0.872[a]
	Knee flexion	50.52 (12.32–56.17)	52.63 (45.63–56.17)	0.143[b]
	Knee external rotation	1.87 ± 6.70	2.18 ± 4.16	0.902[a]
	Knee abduction	5.29 ± 3.74	5.65 ± 2.50	0.802[a]
	Ankle dorse flexion	25.13 ± 4.98	25.01 ± 5.23	0.960[a]
	Ankle inversion	4.92 ± 2.78	6.34 ± 2.27	0.228[a]
	Ankle external rotation	7.23 (−1.47–19.31)	5.79 (1.05–15.71)	0.465[b]
Cartilage thickness (cm)	Medial	2.51 ± 0.52	3.18 ± 0.36	0.004[a,*]
	Lateral	2.58 ± 0.31	3.01 ± 0.58	0.028[a,*]
	Total	2.55 ± 0.38	3.09 ± 0.44	0.008[a,*]

SD, standard deviation; IQR, interquartile range.
[a]*t*-test.
[b]Mann–Whitney *U*-test.
*Denotes a significant difference ($p < 0.05$).

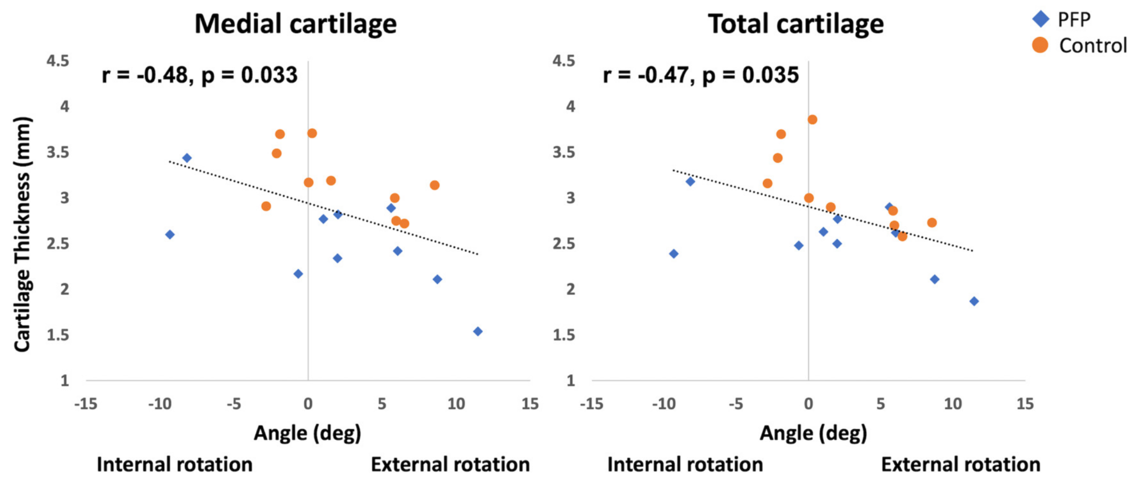

FIGURE 2
Correlations between medial and total patellar cartilage thickness and knee external rotation (KER) in participants with patellofemoral pain (PFP) and pain-free controls.

effect of chronic, abnormal kinematics observed during movement in individuals with PFP (21–23). However, as our study, along with most previous research on cartilage thickness and kinematics in individuals with and without PFP (6, 16), used a cross-sectional design, we were unable to establish causality. Future long-term prospective studies are needed to determine whether greater KER leads to thinner patellar cartilage in individuals with PFP.

4.2 Associations between patellar cartilage thickness and KER in all participants

We found that thinner medial and total cartilage thickness was significantly related to larger KER in all participants with and without PFP. It has been suggested that mechanical stress is a key contributor to the deformation of the PFJ cartilage (24–26). Mechanical stress on the PFJ significantly increases as KER becomes higher in a cadaveric study (11) and in weight-bearing activities in human subject studies (15, 16). Increased KER is believed to elevate the stress on the PFJ due to the lateral tracking of the patella (12, 13). This heightened stress may lead to chronic excessive deformation of the patellar cartilage, ultimately resulting in cartilage thinning (12, 27). We believe that the association between thinner cartilage and greater KER may be due to excessive cartilage deformation from increased stress on the cartilage caused by higher KER. However, as this study is cross-sectional, the causal relationship remains unclear.

Previous studies have reported increased cartilage stress on the lateral facet of the patella with KER (12–14), a finding consistent with our observations in pain-free participants. However, our study found no significant relationship between lateral patellar cartilage thickness and KER in either the entire cohort or specifically in individuals with PFP. Despite this, a trend toward significance was observed (correlation coefficient = 0.394, $p = 0.085$) between lateral patellar cartilage thickness and KER in all participants with and without PFP. This lack of statistical significance may be attributed to the limited range and reduced variability of lateral cartilage thickness in individuals with PFP, potentially reducing the likelihood of detecting a statistically significant correlation. Specifically, we observed an overall thinner lateral cartilage with a smaller standard deviation in the PFP group (mean thickness: PFP = 2.58 ± 0.31 mm; control = 3.18 ± 0.61 mm). To investigate this further, we conducted Levene's test to assess differences in the variance of lateral cartilage thickness between the PFP group and the control group. The test revealed a significant difference in variance (Levene's Statistic = 6.74, $p = 0.018$) between groups, indicating that the PFP group had a smaller variability in the lateral patellar cartilage. These findings suggest that different variability in lateral cartilage thickness between groups may contribute to the observed relationship between lateral cartilage thickness and KER in our study. Therefore, larger-scale studies involving participants with and without PFP and a wider range of PFJ cartilage morphology are needed to examine the associations between patellar cartilage thickness and lower extremity kinematics in greater detail.

4.3 Associations between patellar cartilage thickness and KER in PFP participants

In the PFP group, our study found that thinner medial cartilage was associated with increased KER during bilateral squatting. Theoretically, although stress on the lateral cartilage of the PFJ increases with greater KER (12–14), KER is the strongest predictor of stress across the entire joint cartilage (16). This suggests that greater KER may similarly heighten stress on the medial cartilage, potentially contributing to our observed association between thinner medial cartilage thickness and KER in persons with PFP. Studies highlight that taping interventions aimed at reducing KER significantly decrease both the amount of KER during running (28) and pain (28, 29) in individuals with PFP. These findings suggest a potential link between KER and PFP through its influence on cartilage stress distribution. However, the limited body of causal research on the relationship between medial cartilage stress and KER in PFP underscores the need for further investigation. Furthermore, our study did not find a statistically significant association between lateral or total patellar cartilage thickness and KER during squatting in individuals with PFP. We believe this may be due to the limited range and reduced variability of lateral cartilage thickness in individuals with PFP, as previously discussed.

4.4 Associations between patellar cartilage thickness and KER in pain-free controls

In the control group, consistent with our hypothesis, thinner lateral patellar cartilage was significantly related to greater KER. This suggests that pain-free individuals with greater KER can have thinner lateral cartilage in the PFJ. In addition, thinner lateral patellar cartilage was also significantly related to greater ankle inversion in pain-free participants. Greater ankle inversion has been reported to be associated with increased KER, a factor that raises mechanical stress on the lateral cartilage of the PFJ, in individuals with PFP during gait (30–32). Thus, larger ankle inversion may be linked to thinner lateral cartilage due to this kinematic chain relationship between ankle inversion and KER. On the other side, Reischl et al. (33) have suggested that this relationship between KER and ankle inversion can vary among individuals (33–35). Additional analysis using Spearman's correlation coefficient revealed no significant negative relationship between KER and ankle inversion in the control group ($\rho = -0.60$, $p = 0.067$). Examination of the descriptive statistics revealed that ankle inversion in the control group exhibited a non-normal distribution, with one potential outlier showing an unusually large ankle inversion angle that likely skewed the correlation coefficient. We believe this distributional characteristic might contribute to the observed significant associations between patellar cartilage thickness and ankle inversion. Due to the limited sample size in this study, further research is needed to better understand these relationships.

4.5 Limitations

This study has several limitations. First, as a secondary analysis, we were unable to confirm an ideal sample size, which may have influenced the observed significant relationship between cartilage thickness and KER. Additionally, due to the cross-sectional design and the use of simple correlation analysis, the causal relationship between cartilage thickness and KER remains unknown. In this study, efforts were made to control for potential confounding factors between individuals with and without PFP by including only female participants and matching them for age, weight, height, and activity level. However, as other potential confounding variables influencing KER or patellar cartilage were uncertain, they were not controlled for. Future research should investigate additional confounding factors that may influence outcome measures between the PFP and control groups. Finally, while none of the participants with PFP reported pain during the squatting task, it is important to note that any abnormal kinematics on the painful side may have been compensated for by the asymptomatic limb. Future studies are needed to assess the associations between single-limb weight-bearing activities and PFJ cartilage morphology in persons with and without PFP.

5 Conclusion

This study is, to our knowledge, the first to investigate the relationship between patellar cartilage thickness and KER during bilateral squatting in individuals with and without PFP. Our data showed that thinner medial and total patellar cartilage was associated with increased KER during bilateral squatting in persons with and without PFP. In the PFP group, thinner medial patellar cartilage was associated with increased KER in individuals with PFP. In pain-free controls, thinner lateral and total patellar cartilage was associated with increased KER. These results suggest that increased KER during squatting may negatively affect patellar cartilage thickness, regardless of the presence of PFP. Therefore, clinicians should carefully assess KER during squatting mechanics when evaluating patients with or at risk of PFP, as abnormal movement patterns may serve as a contributor of altered PFJ cartilage morphology. Future research should investigate the influence of lower extremity kinematics, including KER during single-limb activities, on PFJ cartilage morphology to enhance clinical applications.

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 humans were approved by the Health Sciences Institutional Review Board at the University of Southern California. The studies were conducted in accordance with the local legislation and institutional requirements. The participants provided their written informed consent to participate in this study.

Author contributions

HN: Conceptualization, Formal analysis, Visualization, Writing – original draft, Writing – review & editing. K-YH: Data curation, Funding acquisition, Investigation, Methodology, Supervision, Writing – original draft, Writing – review & editing.

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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 tDCS combined with TENS in relieving pain and improving gait patterns during stepping over obstacles among older adults with knee osteoarthritis

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Purpose: Older adults with knee osteoarthritis (KOA) exhibit an elevated risk of falls during obstacle negotiation, primarily due to pain-induced gait deviations. While transcutaneous electrical nerve stimulation (TENS) offers modest pain relief and limited gait modulation, combining it with transcranial direct current stimulation (tDCS) may enhance the effects. This study evaluated the comparative efficacy of tDCS + TENS vs. TENS alone in mitigating pain and optimizing gait patterns during obstacle crossing in older adults with KOA.
Methods: Twenty-three participants with KOA (mean age: 67.6 ± 5.0 years; BMI: 25.9 ± 2.4 kg/m^2) were randomized to either tDCS + TENS ($n = 12$; 7F/5M) or TENS-only ($n = 11$; 7F/4M) groups. Both interventions involved 20-minute sessions, administered thrice weekly for six weeks. Outcome measures included pain intensity (visual analog scale, VAS) and gait variables (foot clearance height, crossing velocity) assessed pre- (week 0) and post-intervention (week 7). Data were analyzed using mixed-design two-way ANOVAs with Bonferroni corrections.
Results: Statistically significant group-by-time interactions were observed for pain ($p = 0.002$, $\eta_p^2 = 0.378$), foot clearance ($p = 0.038$, $\eta_p^2 = 0.190$), and crossing velocity ($p < 0.001$, $\eta_p^2 = 0.588$). *post hoc* analyses revealed that the tDCS + TENS group (week0 = 4.72 ± 1.01, week7 = 1.98 ± 0.88, $p < 0.001$) experienced significantly greater reductions in pain scores compared to the TENS-only group (week0 = 5.02 ± 1.19, week7 = 3.56 ± 1.18, $p < 0.001$); tDCS + TENS group experienced significantly greater improvements in foot clearance (week0 = 0.19 ± 0.04, week7 = 0.20 ± 0.03, $p < 0.001$) and crossing velocity (week0 = 0.53 ± 0.11, week7 = 0.62 ± 0.08, $p < 0.001$), compared to the TENS-only group (week0 = 0.17 ± 0.02, week7 = 0.17 ± 0.02, $p < 0.001$; week0 = 0.52 ± 0.09, week7 = 0.54 ± 0.09).
Conclusion: The combination of tDCS and TENS significantly outperformed TENS-only in reducing pain and enhancing gait adaptability during obstacle negotiation in older adults with KOA. These findings support the integration of tDCS as an adjunctive neuromodulatory strategy to amplify the therapeutic benefits of TENS in this population.

KEYWORDS

knee pain, osteoarthritis, transcranial direct current stimulation, transcutaneous electrical nerve stimulation, obstacle crossing

1 Introduction

Knee osteoarthritis (KOA) is a chronic degenerative joint disorder characterized by progressive damage to articular cartilage, subchondral bone, and the synovial membrane (1). It ranks among the top five causes of disability in older adults (2), with a global prevalence exceeding 645 million individuals (3).

Obstacle negotiation presents distinct challenges for older adults with KOA, as this task exacerbates pain and elevates fall risk. Stepping over obstacles elicits greater pain intensity compared to level walking (4), a hallmark symptom of KOA (5). Approximately 50% of falls in this population occur during obstacle crossing (6), frequently resulting in fractures or mortality (7). Foot clearance, defined as the vertical distance between the foot and the obstacle during the swing phase of gait, is critical for fall avoidance, as most trips occur due to inadvertent contact between the swinging limb and the obstacle. Reduced foot clearance and stepping height is strongly associated with tripping and falls (8). Additionally, patients with KOA exhibit longer single-leg support time and slower gait speeds (9), with a 0.1 m/s decrease in velocity linked to a 10% decline in physical performance capacity (10). These gait alterations may arise from decreased lower limb flexion angles in the leading leg, reduced vertical impulse from the trailing leg during swing initiation, or insufficient propulsive force to maintain gait speed (11, 12).

Standard KOA interventions include pharmacological therapy, surgical procedures, and physical agent modalities (13). Pharmacological analgesics provide transient pain relief but are associated with gastrointestinal and cardiovascular adverse effects (14), while surgical options may be contraindicated in older adults due to comorbidities (15). Physical therapies are favored for their rapid efficacy and safety profiles (16), with transcutaneous electrical nerve stimulation (TENS) being a widely recommended physiotherapy (17). TENS delivers electrical currents via cutaneous electrodes to modulate peripheral pain pathways by activating large-diameter Aβ afferent fibers, which enhance inhibitory interneuronal activity in the dorsal horn of the spinal cord (18). However, TENS exhibits limited analgesic duration and modest effects on functional outcomes (19).

The limited analgesic and functional efficacy of TENS in KOA may be attributed to its exclusively peripheral mechanism of action. According to the classic gate control theory, conventional peripheral physiotherapy (e.g., TENS) modulates pain by activating large-diameter Aβ afferent fibers, which enhance inhibitory interneuronal activity within the dorsal horn of the spinal cord, thereby "closing the gate" to nociceptive input (20). However, the gate control theory also posits a central inhibitory pathway. Aβ fiber signals are rapidly transmitted to the brainstem and cortex, where descending projections modulate spinal gate dynamics via the periaqueductal gray and rostral ventromedial medulla (20). While TENS targets peripheral nociceptive pathways, it fails to directly address central sensitization, a hallmark of chronic KOA pain and disability (21).

Central sensitization in KOA arises from sustained nociceptive input driven by synovial inflammation and sterile inflammation of local soft tissues, which activates peripheral nociceptors and triggers increased neurotransmitter release at primary afferent terminals in the spinal dorsal horn (22). This persistent afferent barrage enhances the responsiveness of nociceptive neurons, leading to heightened pain sensitivity and exaggerated responses to mild stimuli, perpetuating chronic pain (22). Additionally, central sensitization may contribute to functional impairments, including altered gait patterns that further elevate fall risk (23).

Transcranial direct current stimulation (tDCS) is a non-invasive brain stimulation technique that delivers low-intensity direct current via surface electrodes (anode and cathode) positioned over targeted cortical regions (24). By inducing subthreshold shifts in neuronal membrane polarization, tDCS modulates cortical excitability: anodal stimulation enhances excitability, while cathodal stimulation suppresses it. This neuroplastic modulation may attenuate central sensitization-related pain hypersensitivity by activating descending inhibitory pathways in the spinal dorsal horn (25).

Overall, TENS provides transient pain relief via peripheral mechanisms but has limited effects on functional outcomes (19). In contrast, tDCS modulates cortical excitability and central pain processing, potentially improving gait patterns through enhanced neural plasticity (25, 26). A combined tDCS + TENS intervention may synergistically address pain and gait deficits in older adults with KOA by integrating peripheral analgesia with central neuroplasticity enhancement. However, no prior studies have evaluated this approach. Therefore, this study aimed to investigate the effects of a 6-week tDCS combined TENS intervention on pain relief and obstacle-crossing gait improvement (including foot clearance and crossing velocity) among older adults with KOA, compared to TENS alone. We hypothesize that (1) both TENS + tDCS and TENS alone will reduce pain scores and improve obstacle-crossing gait patterns (i.e., increased foot clearance and crossing velocity) in older adults with KOA, and (2) TENS + tDCS intervention will demonstrate superior efficacy compared to TENS alone in reducing pain and enhancing gait adaptability.

2 Materials and methods

2.1 Sample size estimate

An *a priori* power analysis conducted by the G*Power 3.1 software (University of Düsseldorf, Düsseldorf, Germany) indicated that a minimum of 22 participants should be recruited to obtain an alpha level of 0.05 and a statistical power of 0.95 based on a previous study: pain scores decreased more significantly in KOA patients who received 4-week of TENS combined with tDCS interventions compared to those who received only 4-week of TENS combined with sham tDCS interventions with a significant group-by-intervention interaction ($p = 0.038$, $\eta_p^2 = 0.101$) detected in the pain scores using a mixed design two-way ANOVA (27).

2.2 Participants

All participants were recruited from local communities via flyer distribution and presentations from Sep 2024 to Jan 2025. Fifty individuals were screened for eligibility based on the following inclusion criteria: (a) aged 65 years or older; (b) diagnosed with unilateral or bilateral KOA per the American College of Rheumatology clinical criteria (28); (c) Kellgren/Lawrence radiographic grade 2 or 3. Exclusion criteria included: (a) neurological or neuromuscular disorders affecting the knee (other than KOA); (b) history of lower extremity joint surgery or fractures within the past 3 months; (c) planned total knee replacement in the coming months; (d) chronic, disabling back, hip, ankle, or foot pain interfering with daily activities; (e) severe cognitive impairment (Mini-Mental State Examination score <24); (f) intolerance to electrical stimulation (e.g., pacemaker implantation, unusual pinprick sensation).

Twenty-eight eligible participants were randomly allocated (1:1 ratio) to tDCS + TENS or TENS groups using sequentially numbered, opaque, sealed envelopes containing group assignments. The tDCS + TENS group received active tDCS combined with TENS, while the TENS group received sham tDCS combined with TENS, over 6 weeks (three 20-minute sessions weekly). Five participants dropped out by week$_7$, one for relocation and four for poor compliance. Final analysis included 23 participants (12 in the tDCS + TENS group and 11 in the TENS group) (Figure 1). All participants provided written informed consent. The study was approved by the Ethics Committee of Exercise Science, Shandong Sport University (20233037), adhering to the Declaration of Helsinki.

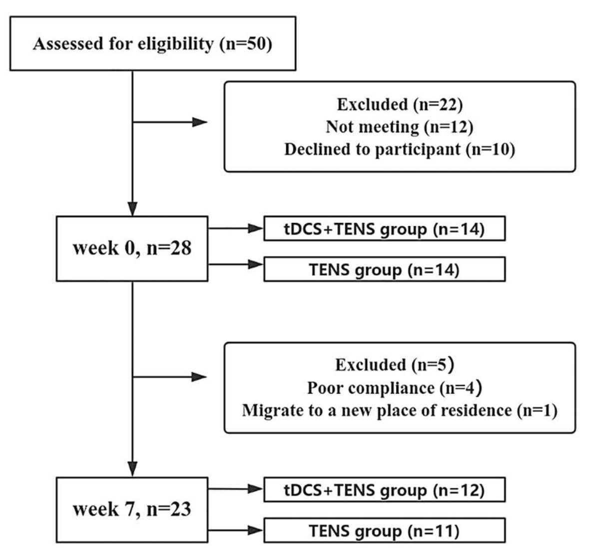

FIGURE 1
Participant flow chart. Participation flow chart from week0 to week7. The final analysis included data from 23 participants. Twenty-seven participants were excluded from the original 50 recruited due to various reasons. tDCS, transcranial direct current stimulation; TENS, transcutaneous electrical nerve stimulation.

2.3 Transcranial direct current stimulation (tDCS) intervention

A tDCS device (Starstim8, Neuroelectronics, Spain) delivered stimulation via two 5 cm diameter rubberized circular electrodes. The anode was precisely positioned over the primary motor cortex (M1) at the Cz electrode site of the 10–20 EEG system. Cz is located at the skull midline, midway between the nasion (nasal root) and inion (external occipital protuberance), corresponding to the lower limb motor cortex. The cathode was placed over the ipsilateral supraorbital (SO) area (FP2 or FP1 of the non-dominant hemisphere), targeting the hemisphere contralateral to the affected knee (determined by higher Kellgren-Lawrence grade or self-reported pain intensity). This formed the M1-SO montage (27) (Figure 2a). Active tDCS delivered a constant current of 2 mA, ramped from 0 mA to 2 mA over 30 s, maintained at 2 mA for 19 min, and tapered to 0 mA over 30 s (total session duration: 20 min). Sham stimulation mirrored electrode placement and initial ramp-up (30 s at 2 mA), followed by immediate shutdown to mimic sensory effects.

2.4 Transcutaneous electrical nerve stimulation (TENS) intervention

Stimulation was delivered via the Low and Medium Frequency Therapy System (Junde Medical Equipment Co., Ltd., Model IN-1300, Hebei, China) using TENS modalities, at the same time when participants received tDCS intervention. Two circular surface electrodes (diameter: 5 cm) were positioned on the medial and lateral sides of the knees, approximately 5 cm apart and centered on the pain site (Figure 2b). Conventional TENS parameters included: 100 Hz frequency, 100 µs pulse width, and a balanced biphasic square waveform. The intensity range of the TENS device is fixed at 0–35 mA and can be continuously adjusted within this range. In this study, the intensities received by the participants were mostly concentrated in the range of 15–25 mA.

2.5 Stepping-over obstacle test

Each participant walked at a self-selected pace on an 8-m walkway and stepped over an obstacle with a height of 20% of each participant's leg length (29). Two force platforms (90*60*10 cm, AMTI, BP600900, USA) were placed adjacent with the long edges and on either side of the obstacle (Figure 3a). The trailing leg steps on the near side of the force platform first, and then the leading leg steps on the far side of the force platform on the other side of the obstacle. Before the tests, the participants were asked to familiarize themselves with the obstacle-stepping process. Forty-three markers were placed on bony landmarks according to the protocol 13-segment whole body model. Three-dimensional kinematics data were collected by a twelve-camera motion analysis system (Vicon, Oxford

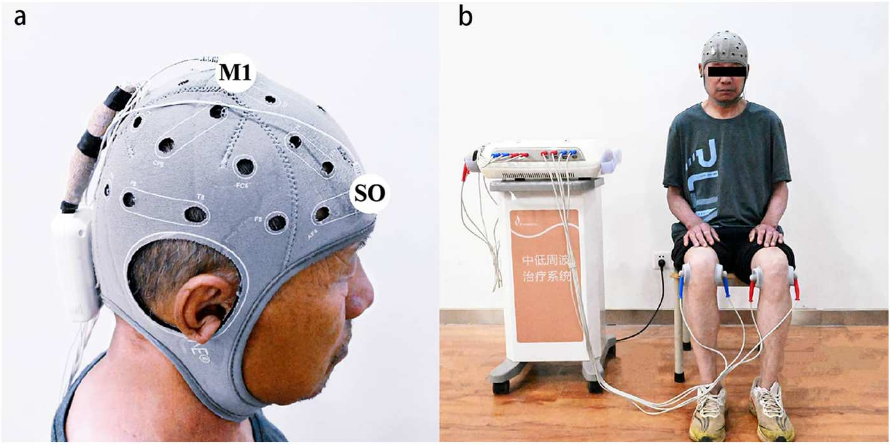

FIGURE 2
Illustration of tDCS electrode and TENS surface electrode placement. **(a)** The illustration of the tDCS electrode placement. The anode electrode was placed over the M1 on the contralateral primary motor cortex (M1) of the affected knee, the cathode electrode was placed over supraorbital (SO) area. **(b)** The illustration of the TENS electrode placement. Two surface electrodes were placed opposite each other on the medial and lateral sides of the knee joints. tDCS, transcranial direct current stimulation; TENS, transcutaneous electrical nerve stimulation.

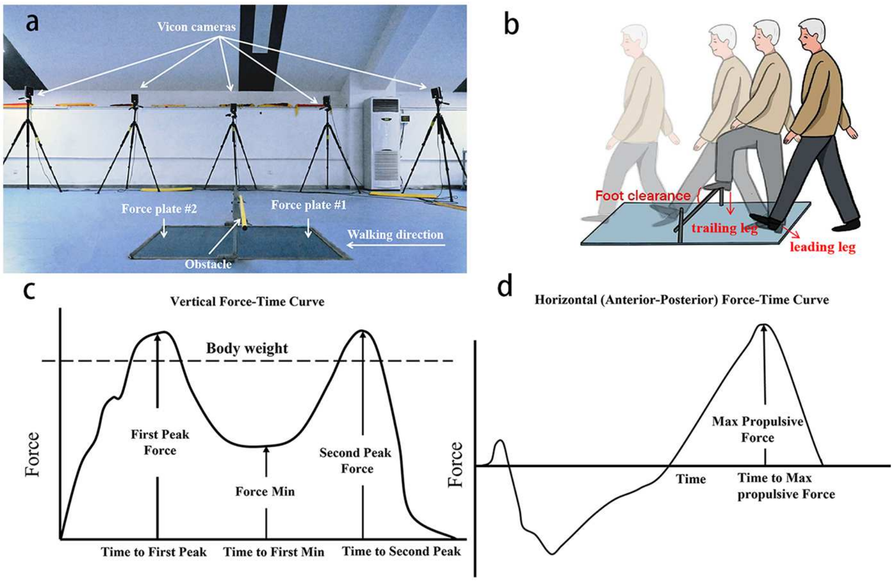

FIGURE 3
Diagram of the obstacle-crossing setup and variables. **(a)** Obstacle and the force platforms setup. **(b)** Diagram of stepping over the obstacle. **(c)** Vertical ground reaction force-time curve. **(d)** Horizontal (anterior–posterior) ground reaction force–time curves.

Metrics Ltd., UK) at 100 Hz. The kinematic data were internally synchronized with the ground reaction force data collected using the force platforms at 1,000 Hz. Each participant was instructed to step over the obstacle using their affected leg (Figure 3b). Three successful trials were collected, a successful trial was defined as a trial the participants used the affected leg as the leading leg and had no contact with the obstacles, and no gait adjustments were adopted during the process.

2.6 Pain scores

Visual analog scale (VAS) was used to assess patient's pain (30). It consists of a 10-cm horizontal line with endpoints labeled "0" (no pain) and "10" (worst possible pain). Participants were instructed to mark their pain level on the line immediately after crossing the obstacle at weeks 0 and 7. Higher scores indicate greater pain intensity.

2.7 Variables

Foot clearance was determined as vertical height between the lowest point on the leading foot and obstacle when the foot was directly over the obstacle (29). Crossing velocity was calculated as the mean anterior-posterior velocity of the center of mass during the stepping-over stride cycle, beginning at the trailing leg heel-strike on the force platform, ending at the next heel-strike of the same leg (31). Hip and knee flexion and ankle dorsiflexion angles were measured as joint Euler angles when the toe marker of the leading leg was directly above the obstacle.

Vertical impulse was computed by integrating the vertical ground reaction force (vGRF)-time curve of the trailing limb from trailing limb heel contact (detected when vGRF >20 N) to trailing limb toe-off (vGRF <20 N). Propulsive impulse was computed by integrating the anterior-posterior ground reaction force (AP GRF)-time curve of the trailing limb during the propulsion phase, defined as the period from the transition to positive AP GRF (forward-directed force) to trailing limb toe-off (vRF <20 N). Stepping height was defined as the maximum vertical distance between the heel marker of the leading limb and the ground surface during the swing phase. Support time was defined as the duration from heel strike of the trailing limb to toe-off of the same limb during obstacle negotiation, measured using force platform data (vertical ground reaction force >20 N threshold).

2.8 Data reduction

Helen Hayes Model in Visual-3D software (C-motion, Germantown, MD) was uesed to process with data. Joint angles were computed via Euler rotations, and hip, knee, and ankle joint centers were determined from marker positions and participant-specific anatomical measurements. Vertical impulse was calculated as the time-integrated vertical GRF during stance phase (Figure 3c), whereas propulsive impulse represented the time integrated anterior-posterior GRF component. GRF were sampled at 1,000 Hz, normalized by body weight (BW) to enable inter-subject comparisons and expressed as a percentage of stance phase duration (Figure 3d). The kinematic and kinetic data were filtered using a fourth-order low-pass Butterworth filter with cutoff frequencies of 6 and 50 Hz (32). Vertical and propulsive impulse was derived from normalized force-time data using trapezoidal-rule integration.

2.9 Statistics

The normality of data was verified using Shapiro–Wilk tests. Mixed-design two-way ANOVAs were used to verify the main effects of group (tDCS + TENS vs. TENS) and time (week0 vs. week7), and their interactions. If a significant interaction was detected, Bonferroni adjusted post hocs would be conducted. Partial eta square(η_p^2) was used to represent the effect size of main effects and interactions. The thresholds for η_p^2 were as follows: 0.01–0.06, small; 0.06–0.14, moderate; >0.14, large (33). Cohen's d was used to represent the effect size of the *post hoc* comparisons. The thresholds for d were as follows: <0.20, trivial; 0.21–0.50, small; 0.51–0.80, medium; >0.81, large (34). The significance level is set to 0.05, and p-value less than the level indicates a statistically significant result, meaning the observed data provide strong evidence against the null hypothesis.

3 Results

All dependent variables exhibited normal distribution, as verified through Shapiro–Wilk tests ($p > 0.05$). Chi-square tests revealed no statistically significant differences in sex ($p = 0.795$) and the side of the more affected leg ($p = 0.795$) between the two groups. Independent t-tests indicated no statistically significant differences in age ($p = 0.828$), height ($p = 0.196$), body mass ($p = 0.055$), and body mass index ($p = 0.078$) between the groups (Table 1).

3.1 Primary outcomes

As shown in Figure 4, significant time × group interactions were detected for pain score ($p = 0.002$, $\eta_p^2 = 0.378$), which decreased in both groups from week$_0$ to week$_7$ (tDCS + TENS: $p < 0.001$, $d = 2.892$; TENS: $p < 0.001$, $d = 1.232$), with greater reductions observed in the tDCS + TENS group compared to the TENS group. Significant time × group interactions were observed in crossing velocity ($p < 0.001$, $\eta_p^2 = 0.588$) and foot clearance ($p = 0.038$, $\eta_p^2 = 0.190$). Crossing velocity (tDCS + TENS: $p < 0.001$, $d = 0.936$; TENS: $p = 0.022$, $d = 0.223$) and foot clearance (tDCS + TENS: $p < 0.001$, $d = 0.283$; TENS: $p = 0.027$ $d = 0.256$) increased in both groups from week$_0$ to week$_7$, with greater improvements observed in the tDCS + TENS group compared to the TENS group.

TABLE 1 Baseline characteristics.

Group	tDCS + TENS group (n = 12)	TENS group (n = 11)	p
Sex	F (7, 58.3%), M (5, 41.7%)	F (7, 63.6%), M (4, 36.4%)	0.795
Affected leg	R (7, 58.3%), L (5, 41.7%)	R (7, 63.6%), L (4, 36.4%)	0.795
Age (y)	67.7 ± 5.0	67.5 ± 5.1	0.828
Height (cm)	159.2 ± 6.8	163.5 ± 10.0	0.196
Body mass (kg)	66.4 ± 6.8	69.1 ± 12.6	0.055
BMI (kg/m^2)	26.2 ± 1.7	25.7 ± 2.8	0.078

Data were presented as mean ± standard deviation. Chi-square tests were used to compare differences in sex, and side of the affected leg. Independent t-tests were used to compare differences in age, height, body mass and BMI between the tDCS + TENS and TENS groups. F, female; M, male; R, right; L, left; tDCS, transcranial direct current stimulation; TENS, transcutaneous electrical nerve stimulation.

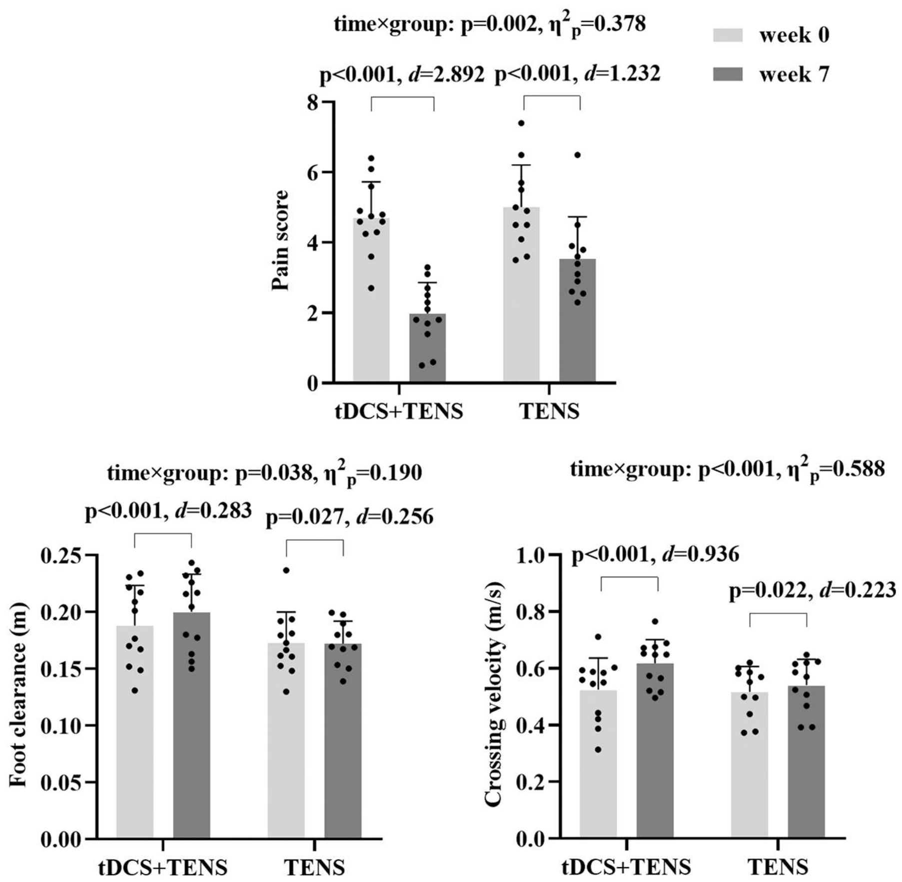

FIGURE 4
Primary outcomes. tDCS, transcranial direct current stimulation; TENS, transcutaneous electrical nerve stimulation.

3.2 Secondary outcomes

As shown in Figure 5. Significant time × group interactions were detected in hip flexion ($p < 0.001$, $\eta^2_p = 0.390$), knee flexion ($p = 0.041$, $\eta^2_p = 0.185$), and ankle dorsiflexion ($p < 0.001$, $\eta^2_p = 0.496$) angles. They increase in both groups from week$_0$ to week$_7$, with the tDCS + TENS group showing greater improvements than the TENS group (tDCS + TENS: hip flexion $p < 0.001$, $d = 1.632$; knee flexion $p < 0.001$, $d = 1.117$; ankle dorsiflexion $p < 0.001$, $d = 1.795$; TENS: hip flexion $p = 0.048$, $d = 0.527$; knee flexion $p = 0.001$, $d = 0.413$; ankle dorsiflexion $p < 0.001$, $d = 1.405$).

Significant time × group interactions were detected for vertical impulse ($p = 0.004$, $\eta^2_p = 0.339$), which decreases in both groups from week$_0$ to week$_7$ (tDCS + TENS: $p < 0.001$, $d = 1.571$; TENS: $p = 0.042$, $d = 0.724$), and greater reductions in the tDCS + TENS group compared to the TENS group. Additionally, a significant main effect of intervention was detected for propulsive impulse in both groups ($p < 0.001$, $\eta^2_p = 0.638$).

Significant time × group interactions were detected for sptepping height ($p = 0.006$, $\eta^2_p = 0.311$) which increased in both groups from week0 to week7 (tDCS + TENS: $p < 0.001$, $d = 1.897$; TENS: $p < 0.001$, $d = 1$), with greater improvements observed in the tDCS + TENS group compared to the TENS group. Significant time × group interactions were detected for support time ($p = 0.004$, $\eta^2_p = 0.334$) which increased in both groups from week0 to week7 (tDCS + TENS: $p < 0.001$, $d = 1.307$; TENS: $p < 0.001$, $d = 0.767$, with greater reductions observed in the tDCS + TENS group compared to the TENS group.

4 Discussion

The purpose of this study was to verify the effect of tDCS combined with TENS on pain and gait patterns during stepping over obstacle among older adults with KOA. These results supported hypotheses # 1 and 2, by pointing out that both interventions relieve pain and improving gait patterns during stepping over obstacles among older adults with KOA, while tDCS + TENS training has better effects.

Our study showed that both interventions were effective in reducing pain, while tDCS + TENS had better effects in older adults with KOA. The finding is supported by a previous study, which indicated that combination of tDCS and TENS is more

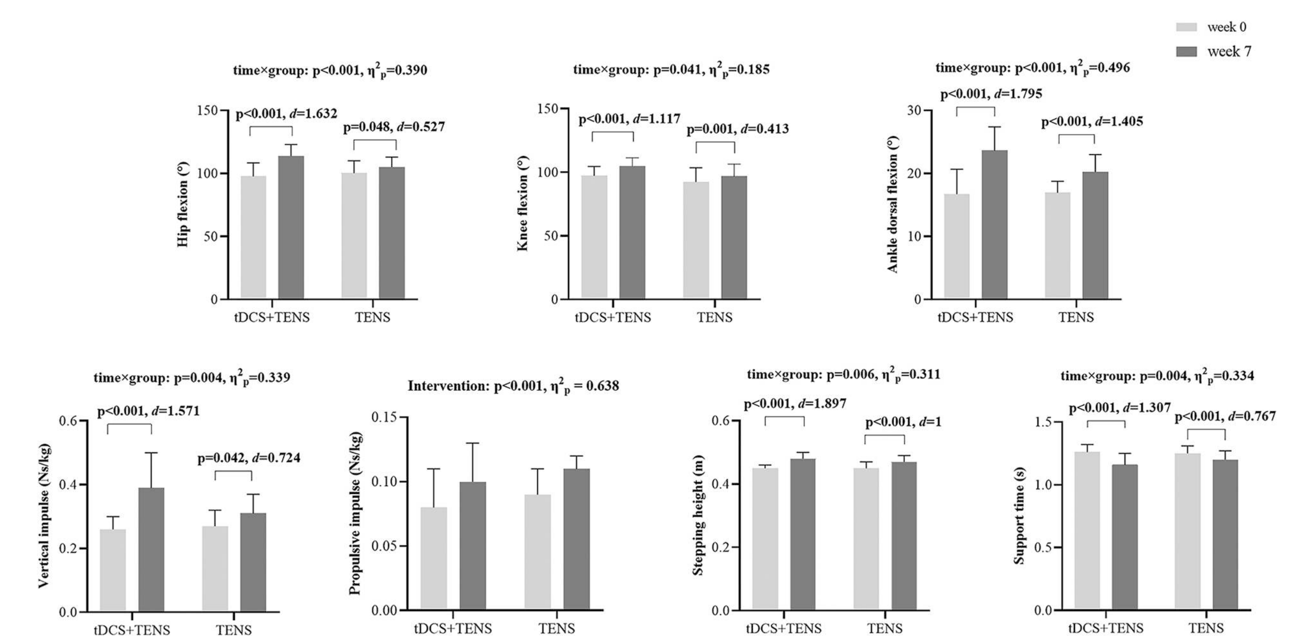

FIGURE 5
Secondary outcomes. tDCS, transcranial direct current stimulation; TENS, transcutaneous electrical nerve stimulation.

effective in relieving pain for individuals with KOA than using TENS or tDCS only (35). According to the classic gate control theory, pain perception is regulated by a gating mechanism in the spinal cord. In the dorsal horn, central transmitting cells, known as T-cells (transmission cells), act as relays for nociceptive signals. Glial cells and inhibitory interneurons function as a "gate" by modulating input to T-cells. This gate is regulated by signals from large-diameter and small-diameter fibers, the former closes the gate while the later opens the gate. TENS works by activating large-fiber fibers, thereby enhancing the inhibitory effect of interneurons and closing the gate to block peripheral nociceptive input (20, 36). However, in chronic KOA, persistent peripheral inflammation may lead to central sensitization. tDCS, in contrast, promotes neural plasticity by modulating thalamocortical circuits and synaptic plasticity, which reduces abnormal neuronal discharges associated with central sensitization and alleviates chronic pain-related hyperalgesia (37). TENS reduces peripheral nociceptive input, while tDCS suppresses central pain amplification. Together, these interventions synergistically attenuate pain by targeting both peripheral and central mechanisms.

Our study demonstrated that both interventions were effective in increasing foot clearance during obstacle-stepping in older adults with KOA, while tDCS + TENS yielded superior effects to TENS alone, likely due to two factors. Firstly, adequate toe clearance relies on sufficient hip and knee flexion and ankle dorsiflexion (38); our secondary outcomes revealed that combined tDCS and TENS significantly increased these joint angles during obstacle-stepping in this population, supported by prior research showing both TENS and tDCS can improve joint range of motion during functional tasks (39). Secondly, adequate toe clearance may also stem from higher vertical impulse in the trailing leg and increased spanning height, as our secondary outcomes indicated that combined tDCS + TENS outperformed TENS alone in this regard. During obstacle-stepping, limb control depends on joint movement; TENS alleviates knee pain via spinal gating mechanisms (Aβ-fiber activation) and endogenous β-endorphin release, which reduces pain-induced inhibition of quadriceps and gastrocnemius activation (40). This improves knee flexion and ankle dorsiflexion during the swing phase while simultaneously strengthening push-off forces and increasing spanning height through enhanced muscle activation. The generated greater vertical impulse directly results in improved foot clearance during obstacle-stepping. tDCS effects target central neural networks to amplify motor control and propulsion efficiency. Anodal tDCS applied over the primary motor cortex (M1) contralateral to the affected knee enhances corticospinal tract excitability (41), increasing the firing rate of pyramidal neurons and improving the synchronization of motor unit recruitment in lower limb muscles. This heightened cortical drive optimizes joint movement control, enabling more precise regulation of hip, knee, and ankle angles, strengthening push-off forces and increasing spanning height, ultimately enhancing foot clearance. Together, these interventions synergistically amplifies both joint mobility and propulsion efficiency, thereby achieving greater foot clearance during obstacle-stepping. tDCS enhances neuronal firing rates by boosting excitability in intracortical and subcortical networks and promoting synaptic plasticity (42), thereby improving joint movement control and strengthening push-off forces, ultimately enhancing foot clearance. Together, these interventions synergistically amplifies both joint mobility and propulsion efficiency, thereby achieving greater foot clearance during obstacle-stepping.

Our study demonstrated both interventions were effective in increasing obstacle crossing velocity in older adults with KOA, while the tDCS combined TENS is more effective than TENS alone. The support time and time main effect of propulsive impulse from the secondary outcomes supported both tDCS + TENS and TENS increases crossing velocity. Firstly, tDCS improved motor unit synchronization in the plantarflexors, increasing the rate of force development during push-off (41). This allowed the support leg to generate sufficient propulsion in less time, shortening support duration without compromising stability. TENS reduced pain-related muscle inhibition, allowing the quadriceps and gluteals to better stabilize the knee and hip during support, reducing the need for prolonged ground contact (40). Sencondly, tDCS induces cortical depolarization in the primary motor cortex (M1), increasing corticospinal tract excitability (41). This neuromodulatory effect refines the coupling between cortical motor commands and peripheral muscle activation, minimizing delays between neural input and mechanical output. This effect improves motor unit synchronization of plantarflexor muscles, enhancing push-off force generation during terminal stance phase. TENS activates Aβ-fiber mediated spinal gating and promotes β-endorphin release. These mechanisms reduce in pain-related muscle inhibition directly enhance joint kinematics during propulsion (40). By relieving quadriceps inhibition, TENS enables fuller knee extension during late stance, a movement that increases the leverage of the support leg. This extended knee position shifts the body's COM forward relative to the support foot, amplifying the mechanical advantage for generating forward momentum (11). Simultaneously, reduced inhibition of plantarflexors promotes greater ankle plantarflexion during push-off, which augments the force of propulsion (12). However, the lack of time by group interaction in propulsive impulse indicated that there are other factors attribute to the superior effects of tDCS combined TENS intervention, which may be attribute to the two factors. Firstly, tDCS augments proprioceptive signal integration through anodal stimulation of the primary motor cortex (M1), which increases neuronal excitability in this region, thereby improving the processing of proprioceptive inputs from muscles and joints (e.g., signals from muscle spindles and Golgi tendon organs) (43). This enhanced integration refines real-time perception of limb position and movement dynamics, enabling patients to rapidly adjust foot trajectory and joint angles during stepping. By minimizing positional errors that typically induce deceleration or hesitation, the intervention optimizes movement efficiency, ultimately elevating overall crossing speed.Secondly, tDCS applied over the primary motor cortex (M1) enhances multi-joint coordination through synchronized activation of synergistic neural networks across the ankle, knee, and hip joints by modulating corticospinal excitability and thalamocortical connectivity (44). This optimization reduces kinematic delays during obstacle negotiation, such as inadequate toe clearance or excessive swing-leg flexion, leading to smoother gait transitions and increased overall speed.

This study has limitations. First, there was no follow-up after the 6-week intervention; it cannot be determined how long the effect of the intervention on relieving pain and improving gait patterns during stepping over obstacles in older adults with KOA lasted. Second, the obstacle height was set at only 20% of leg length; incorporating additional heights could enhance the generalizability of the findings. Third, this study included older adults with both unilateral and bilateral KOA, but the gait patterns for stepping over obstacles may differ. Additionally, this study highlights the temporal aspects of ground reaction forces during gait but doesn't explore how individual joints contribute to gait pattern differences. Future research should integrate kinematic and kinetic data to better understand biomechanical changes in KOA patients. Machine learning, which has been effective in identifying key gait features and subtle movement differences (45, 46), could be particularly useful. Using these techniques may help clarify each joint's role in gait variability, improving diagnostic accuracy and guiding targeted interventions.

5 Conclusion

Both the combination of tDCS and TENS and TENS-only were effective in relieving pain and improving gait patterns during obstacle crossing in older adults with KOA, while the combination of tDCS and TENS had superior efficacy. These findings support the integration of tDCS as an adjunctive neuromodulatory strategy to amplify the therapeutic benefits of TENS 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 humans were approved by the Ethics Committee of Exercise Science, Shandong Sport University. The studies were conducted in accordance with the local legislation and institutional requirements. The 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

XZ: Data curation, Writing – original draft. DW: Data curation, Formal analysis, Writing – review & editing.

QinS: Data curation, Formal analysis, Writing – review & editing. XL: Writing – review & editing. YG: Writing – review & editing. PS: Data curation, Methodology, Writing – review & editing. QipS: Conceptualization, Funding acquisition, Methodology, Project administration, Supervision, Writing – review & editing.

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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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Does body weight support improve neural and biomechanical measures during treadmill gait in children with unilateral cerebral palsy?

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Introduction: Body weight support (BWS) treadmill training, commonly utilized to improve gait, has inconsistent evidence of effectiveness across disorders.
Methods: We aimed to comprehensively evaluate its scientific rationale by comparing immediate effects of two weight support levels (20%, 40%) to unsupported (0%) treadmill walking on neural and biomechanical measures in children with unilateral cerebral palsy (CP) and typical development (TD). We hypothesized BWS would demonstrate positive effects only in CP. Participants included 10 with TD and 8 with CP (mean age = 14.6 and 15.4 years, respectively).
Results: Minimal or no group differences or BWS effects were found for synergy number, structure or Walk-DMC, whereas the Gait Deviation Index (GDI) showed a significant interaction with 20% BWS where the dominant side in CP improved with 20% BWS while both sides in TD worsened. Beta band EEG activation from 0% to 20% BWS showed a significant triple interaction increasing in the non-dominant and decreasing in the dominant hemisphere in TD, while increasing in both in CP. A worsening trend was seen with 40% BWS in all measures except z scores.
Conclusion: BWS has beneficial effects on kinematics in CP supporting the basic premise for use in neurorehabilitation at the body structure level.

KEYWORDS
muscle synergies, electroencephalography, kinematics, temporal-spatial, non-negative matrix factorization, unloading

1 Introduction

Cerebral Palsy (CP), which affects approximately 1 in 345 children in the U.S. (1), is a heterogeneous group of neurological disorders due to injuries or insults to the developing brain specifically disrupting motor as well as other aspects of development (2, 3). Walking ability, often a major concern for families of children with CP, may be limited by spasticity, dystonia, poor motor control, muscle weakness, and secondary musculoskeletal changes (4). Mobility levels in CP, as described by the Gross Motor Function Classification System (GMFCS) (5), range from walking independently with only mild incoordination to being dependent on others for all mobility. Many interventions aim to improve mobility in CP to promote greater functional independence and participation in everyday life. Surgical options include tendon lengthening and selective dorsal rhizotomy whereas non-surgical options include targeted muscle chemo denervation, oral anti-spasticity medications, and intensive task-specific physical therapy training, e.g., partial body weight supported treadmill training (6, 7), which is the focus of the current study.

1.1 Review of related work

The neurophysiological basis for body weight support (BWS) treadmill training (TT) was established by studies in spinal-lesioned cats, subsequently translated to humans, demonstrating that this type of intervention involving task-oriented training and progressive practice could promote locomotor recovery (8, 9). A harness was required to provide postural support when no longer present or severely impaired and it was further suggested that decreasing the load on the lower limbs would facilitate step initiation during treadmill walking. Despite little scientific evidence to support its efficacy or effectiveness, BWS TT was rapidly adopted by the rehabilitation community to improve mobility in multiple neurological disorders. A large NIH-funded randomized clinical trial, the Locomotor Experience Applied Post-Stroke or LEAPS) Trial was conducted on over 400 individuals 2 or 6 months post-stroke (10). It compared early or later BWS TT to an equal amount of time with a therapist working on functional mobility. Unexpectedly, neither BWS TT group had superior outcomes compared to standard care, leading many to challenge its implementation in neurorehabilitation (11) with the evidence still insufficient to support this in individuals with disorders such as stroke, spinal cord injury and traumatic brain injury (12). While some of the earlier studies in CP, including randomized controlled trials, had also failed to show that BWSTT is superior to overground walking or other effective mobility training approaches (6, 13, 14), a more recent evidence synthesis (7) now lists this among the effective strategies for functional mobility improvement in this population, which is difficult to reconcile given the results in other populations.

In clinical and research settings, the amount of body weight support may range from 0% to 50% (15–17), and this factor alone may contribute to differences in outcomes across studies. Unloading levels have largely been determined through subjective visual inspection of gait patterns that most closely mimic natural walking characteristics (18). Therefore, more rigorous and comprehensive investigations are warranted to elucidate peripheral and central nervous system neuromuscular control mechanisms and biomechanical adaptations across different body weight support levels, thereby providing evidence to establish objective parameters for optimizing BWS protocols in clinical practice.

The present study investigates the effects of BWS during treadmill walking by comparing individuals with unilateral CP to an age-matched cohort with typical development (TD) across three BWS conditions (0%, 20%, and 40%). An analysis of muscle synergies, defined as groups of muscles that are recruited and activated as a single unit (19–23), was utilized as a primary outcome of neural control, including synergy number, Walk-DMC (dynamic motor control index during walking), and quantitative measures of group- and individual-specific synergy structures based on clustering analysis and z-score distributions. Previous research showed that fewer synergies and a lower Walk-DMC index in CP, indicated reduced motor command complexity when compared with typically developing children (24). Distinct differences in muscle synergy structures between CP and typically developing children have also been observed during unsupported treadmill walking, suggesting altered neuromuscular coordination strategies in CP (25, 26). Additionally, while primarily demonstrated in healthy adults, electroencephalographic (EEG) analysis has demonstrated that muscle synergy activation patterns during walking can be successfully decoded from cortical signals (27). Thus, we also included measures of cortical activation patterns in motor-related brain regions using EEG, along with biomechanical outcomes such as temporal-spatial gait parameters and the gait deviation index (GDI) (28).

1.2 Hypotheses

Based on the assumption that unweighting improves kinematic patterns in those with gait abnormalities, we anticipated that synergy numbers and Walk-DMC values in CP would increase with BWS along with related improvements in gait parameters, and that their synergy structures and EEG patterns would become more similar to those more commonly seen in TD. In contrast, we anticipated that unweighting was not likely to improve gait in TD, and therefore, outcomes would remain basically unchanged or worsened in that cohort.

2 Materials and methods

2.1 Participants

The initial group of participants included 9 children with unilateral CP (7 females, 2 males) and 10 children with TD (8 females, 2 males) (Table 1). The greater numbers of females with CP recruited was not by design, since participants of both sexes were welcome to participate. However, to control for any possible differences in these data with respect to sex, although there is no evidence to suspect sex differences except for shorter step lengths in females due to height differences, we recruited controls to match the make-up of the group with CP. Participants ranged in age from 7 to 21 years with a mean of 15.4 years in CP and 14.6 for TD (p – 0.61). Height and weight were slightly but not significantly greater in TD ($p = 0.77$ and 0.44, respectively). Of the 9 children with unilateral CP, 5 were GMFCS Level I and 4 were GMFCS Level II (Table 1) which are

TABLE 1 Participant characteristics for those with cerebral palsy (CP) and typical development (TD).

Group ID	Age (yrs)	Height (cm)	Weight (kg)	Handedness	Gender	GMFCS
CP1	14	162	43.7	Right	Female	I
CP2	21	166	54.0	Left	Female	II
CP3	16	180	89.2	Left	Male	I
CP4	17	161	75.7	Right	Female	I
CP5	17	178	63.3	Right	Male	I
CP6	13	156	51.1	Left	Female	II
CP7	17	156	57.2	Left	Female	I
CP8	7	120	18	Right	Female	II
CP9	17	174	82.9	Left	Female	II
CP mean	**15.44**	**161.44**	**59.46**	–	–	–
TD1	14	167	81.6	Right	Female	–
TD2	16	165	56.4	Right	Female	–
TD3	18	166	62.8	Right	Female	–
TD4	16	171	92.4	Right	Male	–
TD5	14	154	50.5	Right	Female	–
TD6	18	177	100.4	Right	Female	–
TD7	15	164	65.9	Right	Female	–
TD8	13	168	63.9	Right	Female	–
TD9	7	123	20.8	Right	Female	–
TD10	15	183	81.1	Right	Male	–
TD mean	**14.6**	**163.8**	**67.6**	–	–	–
p value	0.61	0.77	0.44			

Bold indicates $p < 0.05$.

the two highest out of five mobility levels and indicate that all were able to ambulate independently overground without any mobility aids. This was important because to participate in the study, all had to be able to walk on the treadmill without holding onto the handrails. All participants under 18 and their legal guardians or those over 18 provided informed assent and consent, as appropriate. This protocol was approved by the National Institutes of Health Institutional Review Board (Protocol # 13-CC-0110). Only those who could walk on the treadmill without needing to use arm support were included.

2.2 Procedures

The experimental protocol consisted of three treadmill walking conditions (0%, 20%, and 40% BWS). Participants initially were instructed to stand still for 2 min to obtain a resting baseline. Then participants walked for 5 min first at a self-selected speed with 0%, then 20% and 40% BWS in a randomized order using the Zero-G harness which maintains a consistent level of unweighting regardless of the vertical position of the center of mass. If a participant was not able to maintain their walking pace or needed to use arm support during a walking condition, the trial was discontinued and recorded as missing data.

Neural and biomechanical measures were collected using three synchronized measurement systems (EEG, EMG, and motion analysis). A 64-channel, wireless, active EEG system (Brain Products, Morrisville NJ) was positioned within a snugly fitted EEG cap on the participant's head and electrodes were placed according to the International 10/20 system with data collected at 1,000 Hz. EMG data were recorded wirelessly (Trigno Wireless, Delsys, Boston, MA, United States) at 1,000 Hz from surface electrodes positioned on the skin over the right and left eight muscle bellies of the tibialis anterior (TA), medial gastrocnemius (MG), soleus (SOL), peroneus longus (PL), rectus femoris (RF), vastus lateralis (VL), medial hamstrings (MH), and hallicus longus (HL). A 10-camera motion capture system (Vicon, Lake Forest, CA) collected kinematic data at 100 Hz. Reflective markers were placed on anatomical landmarks to track the position of the feet, shank, thigh, and pelvis segments. All kinematic data were processed using Visual3D software (C-Motion, Germantown, MD, USA). These data were utilized to compute temporal-spatial gait parameters (i.e., gait speed, cadence and step distance) and to calculate the Gait Deviation Index (GDI). The GDI is a single number that is a validated indicator of the overall degree of gait pathology compared to a normative reference group based on 15 selected kinematic features, as detailed in (28). A GDI score of 100 indicates normal gait with each 10-point increment below that representing one standard deviation from normal. In this study, we utilized gait data from our 10 participants with TD to serve as the reference group.

EMG data were processed using 35 Hz high-pass and 5 Hz low-pass Butterworth filters (6th order). For synergy analysis, EMG data were divided into 20-cycle windows with a one-cycle sliding window and normalized to maximum activation within each window, as previously described (29). Non-negative Matrix Factorization (NMF) was applied to extract muscle synergies from the processed EMG matrices (EMGo) (30): $\mathrm{EMG}_0 = \sum_{i=1}^{n} W_i C_i + e$, $\mathrm{EMG}_r = \sum_{i=1}^{n} W_i C_i$ where n is the number of synergies from 1 to 16, i is an identification number of each synergy, W represents synergy structure weight vector,

C indicates activation coefficients, and e is residual error. EMGr is a reconstructed EMG matrix resulting from the multiplication of W and C.

The number of synergies was determined using a 90% variability accounted for (VAF) threshold as follows: $\mathrm{VAF} = 1 - (\mathrm{EMG}_o - \mathrm{EMG}_r)^2/\mathrm{EMG}_o^2$. Based on VAF, the dynamic motor control index during walking (Walk-DMC) was computed to quantify neuromuscular control complexity (24): $\mathrm{walk-DMC} = 100 * 10\left[\frac{\mathrm{VAF}_{\mathrm{AVE}} - \mathrm{VAF}_1}{\mathrm{VAF}_{\mathrm{STD}}}\right]$, where VAF1 represents the VAF with one synergy, and VAFAVE and VAFSTD are the mean and standard deviation of VAF1 across the typically developing cohort. Higher VAF1 resulting in lower Walk-DMC indicates simplified neuromuscular control.

To identify group-specific characteristics under different BWS conditions, muscle synergy structures were classified using k-means clustering and discriminant analyses as detailed in (29). Then, clusters were categorized as CP-specific (C), TD-specific (T), or non-specific based on two-proportion z-test results ($z > 1.96$, $z < -1.96$, or $-1.96 \le z \le 1.96$, respectively). Within each category, clusters were numbered according to their z-values (e.g., C1 representing the CP-specific cluster with highest z-value). To quantitatively assess how much each participant exhibited group-specific muscle synergies, individual weight-averaged z scores were computed from the number of synergies and corresponding cluster z-values, with positive scores indicating stronger CP representation and negative scores indicating stronger TD representation. This analysis was performed separately for each walking condition and averaged across conditions.

2.3 EEG processing

EEG data analysis procedures were based on our prior studies (25, 31) and are summarized in Figure 1. EEG and motion capture data were synchronized to align gait events with EEG data. EEG data were then high pass filtered at 1 Hz and noisy channels and time periods were removed before downsampling and then concatenating across conditions (standing, walking with and without 20% BWS). Next, an artifact subspace reconstruction (ASR) algorithm (32) was used to identify and reconstruct time periods corrupted by non-stereotypical artifact. The ASR-cleaned data were then common average referenced and an adaptive mixture independent component analysis (AMICA) (33) was used to extract independent sources (ICs) from EEG. The AMICA transformation was then applied to the downsampled, non-ASR cleaned data sets. Next, equivalent dipoles were fit to each IC using the DIPFIT toolbox in EEGLAB with a template, 3 shell boundary element head model. ICs with dipole fits having greater than 20% residual variance (RV) (31) and topographical sparseness (TS) less than 5 (34) were rejected. IC power spectra, scalp topography, and dipole location were also visually inspected to remove non-cortical sources such as eye blink, EMG, etc. Dipole locations were then adjusted based on hand dominance, such that the left hemisphere (right hand) was represented as the dominant side for all participants.

Next, individual strides from consecutive heel-strikes were extracted as were non-overlapping 2 s epochs from the standing baseline data. K-means was then used for clustering to pool ICs from both groups across all conditions using their dipole coordinates, power spectral density (PSD), and inter-trial coherence (ITC) as coordinates in the parameter space, where PSD and ITC were reduced to 3 dimensions using principal component analysis (PCA). PSD and ITC were equally weighted in the k-means algorithm while the dipole coordinates were assigned a weight three times greater. Two sensorimotor-related clusters were identified, one in each hemisphere, based on the centroid location and confirmed by the presence of mu and beta event-related desynchronization (ERD) during walking relative to standing. The event related spectral perturbations (ESRPs) of each gait cycle were computed to quantify the de-synchronization or synchronization at each time point, by dividing the walking power spectra by the respective standing (rest) mean spectra for each IC. The ICs within each cluster

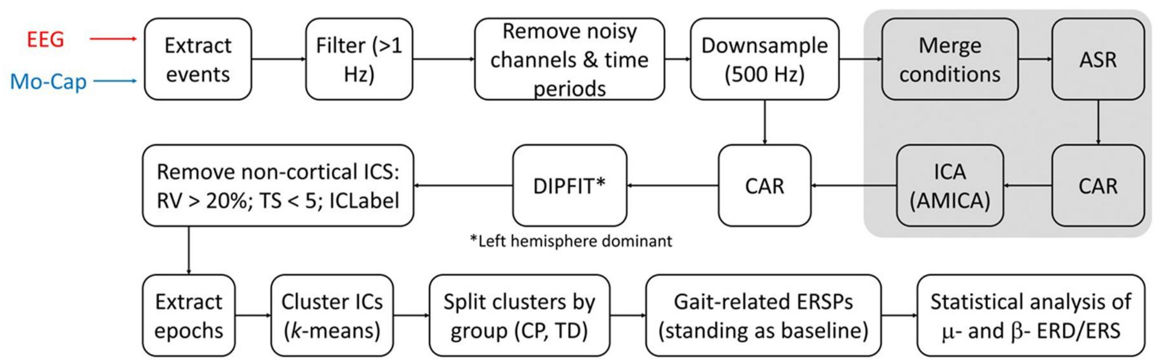

FIGURE 1
Processing steps for analyzing the differences in motor-related cortical activity between groups (typical development [TD], cerebral palsy [CP]) and walking conditions (with and without 20% BWS) from scalp-recorded EEG.

were split by group (TD, CP) and the overall group mean ERSP for each condition was computed. Finally, statistically significant differences in the mean mu (8–13 Hz) and beta (15–30 Hz) ERSPs for each cluster were computed across conditions and groups using nonparametric bootstrapping via the condstat function in the EEGLAB using 2,000 points of surrogate data with the significance (alpha) set to 0.05.

2.4 Statistical analysis

Statistical analyses for synergy number, z scores, temporal-spatial and kinematic gait data, and EEG ERD/ERS values were performed using a general linear mixed model in SPSS (version 31.0 using the full factorial default model with polynomial contrasts) with group (CP or TD) as the between subject variable and BWS as the within subject variable, first with all who had complete data for 0% and 20% BWS and then for all who had complete data for 0, 20% and 40% BWS. The statistical significance was set at $p < 0.05$. *Post hoc* tests were performed as indicated. Baseline data as well as change scores were correlated across outcome categories using Pearson r for the entire sample and those with CP separately.

3 Results

3.1 Participants

While data from all participants and conditions were used to determine the clusters, statistical comparisons of conditions, limbs and groups only included participants able to perform one or both BWS conditions. Of the 9 with CP and 10 with TD who were able to perform the no BWS conditions, one participant (CP8) could not perform either the 20% or 40% BWS condition without holding onto the handrails; therefore, was only included in the clustering analysis. Additionally, two participants with CP (CP3 and CP4) were not able to complete the 40% BWS condition for the full duration without holding the handrails. For two other participants with CP (CP1 and CP2) and two with TD (TD1 and TD10), the BWS harness moved vertically up and down during the 40% condition, resulting in discomfort and inaccurate motion capture data; therefore, these participants were excluded from the 40% BWS analysis. Finally, a technical error in the synchronization signal between EMG, EEG and motion capture occurred during 40% BWS condition in TD2 who was excluded as well. To test the assumption that those who could not perform or were excluded from the highest BWS support condition were less functional due to age or other factors, we compared participants who could not perform the 40% BWS condition with those who could. Interestingly, age and baseline temporal-spatial or GDI data did not differ significantly between these subgroups; synergy numbers also did not differ (see Table 2 a & b; all

TABLE 2 Mean outcomes for groups with typical development (TD) and cerebral palsy (CP) for each body weight support (BWS) condition.

Outcomes	*n* for TD/CP	TD	CP
Synergy number			
Baseline	10/8	5.30 (0.48)	5.25 (0.46)
20% BWS	10/8	5.20 (0.63)	5.25 (0.46)
40% BWS	7/4	5.00 (0.00)	4.75 (0.50)
Walk-DMC			
Baseline	10/8	100.0 (10.0)	93.1 (9.61)
20% BWS	10/8	100.0 (10.0)	98.4 (14.82)
40%BWS	7/4	100.0 (10.0)	94.2 (8,04)
Synergy *z* score			
Baseline	10/8	−16.4 (5.68)	12.0 (11.25)
20% BWS	10/8	−18.9 (7.63)	12.0 (10.51)
40%BWS	7/4	−14.3 (6.12)	13.5 (7.64)
Gait speed (m/s)			
Baseline	10/8	0.99 (0.11)	0.89 (0.10)
20% BWS	10/8	0.99 (0.11)	0.89 (0.10)
40%BWS	7/4	0.99 (0.11)	0.89 (0.11)
Cadence			
Baseline	10/8	104.3 (5.52)	102.4 (10.7)
20% BWS	10/8	102.7 (5.70)	101.4 (10.5)
40%BWS	7/4	100.1 (7.2)	96.3 (10.6)
Dominant step distance			
Baseline	10/8	0.51 (0.05)	0.50 (0,03)
20% BWS	10/8	0.51 (0.06)	0.49 (0.03)
40%BWS	7/4	0.51 (0.08)	0.47 (0.09)
Non-dominant step distance			
Baseline	10/8	0.50 (0.07)	0.48 (0.04)
20% BWS	10/8	0.51 (0.07)	0.49 (0.05)
40%BWS	7/4	0.51 (0.09)	0.48 (0.05)
Gait deviation index - dominant			
Baseline	10/8	95.7 (10.8)	77.8 (10.8)
20% BWS	10/8	91.0 (10.5)	82.0 (10.0)
Gait deviation index – non-dominant			
Baseline	10/8	95.8 (8.40)	72.4 (10.8)
20% BWS	10/8	91.9 (8.83)	73.4 (6.26)
Mu ERD – dominant			
Baseline	9/7	−4.76 (1.52)	−4.79 (3.45)
20% BWS	9/7	−4.61 (1.24)	−5.36 (4.14)
Mu ERD – non-dominant			
Baseline	8/5	−3.64 (1.29)	−4.84 (1.36)
20% BWS	8/5	−3.93 (1.56)	−5.38 (2.28)
Beta ERD – dominant			
Baseline	9/7	−2.96 (0.78)	−3.26 (1.13)
20% BWS	9/7	−2.84 (0.64)	−3.64 (1.00)
Beta ERD – non-dominant			
Baseline	8/5	−2.70 (0.64)	−3.10 (0.33)
20% BWS	8/5	−3.03 (0.30)	−3.19 (0.72)
Z score			
Mean	10/8	−16.4 (4.5)	12.1 (9.5)
Baseline	10/8	−17.4 (5.2)	11.0 (10.2)
20% BWS	10/8	−17.0 (6.0)	13.3 (8.8)
40% BWS	7/4	−14.6 (6.1)	12.5 (7.4)

Note that values for 40% BWS are not directly comparable to the larger sample for 0 and 20% BWS.

p values > 0.80). The distribution of GMFCS levels in CP was similar in those who could and could not perform the 40% BWS condition.

3.2 Muscle synergy results

Across all conditions, the mean number of gait cycles used for the analyses was 216.1 ± 35.5 for the group with TD and 214.5 ± 46.9 for the group with CP which did not differ across groups ($p = 0.91$). The GLM analysis revealed no significant group differences in synergy number in any condition and no interactions for the 0% and 20% BWS (BWS condition $p = 0.71$; group $p = 0.71$; group by condition interaction $p = 1.00$) and 0, 20% and 40% BWS cohorts (BWS condition $p = 0.10$; group $p = 0.80$; interaction $p = 0.82$), indicating that the group with CP did not demonstrate baseline differences in synergy number, nor did BWS significantly alter synergy numbers in either group (see Table 2 for values of all measures by group and condition).

While the mean walk-DMC for the group with CP was lower than 100, the GLM results comparing the 0% and 20% BWS conditions did not demonstrate any significant condition or group differences or interactions ($p = 0.19$ for BWS, $p = 0.19$ for interaction, $p = 0.40$ for group). Interestingly, there was a significant difference (worsening) for the BWS condition ($p = 0.04$) when comparing 20% and 40% in the whole sample ($p = 0.41$ for interaction, $p = 0.97$ for group). The mean value for the group with CP moved closer to the normative TD Walk-DMC value of 100 with 20% BWS but further away with 40% BWS.

3.2.1 Clustering: muscle synergy

Nineteen distinct clusters across all participants and conditions were identified. Muscle activation patterns for each cluster are shown in Supplementary Figure S1. To determine whether a cluster was specific to the TD or CP group, the proportion of synergies for each cluster was calculated using the two-proportion z-test. Based on these values, 12 CP-specific clusters and 7 TD-specific clusters were identified (Figure 2). Of the 12 CP clusters, 6 were present in both groups (C1, 3, 4, 6, 9, 12) whereas the rest were present only in CP (C2, 5, 7, 8, 10, 11). Clusters C8, C10, and C11 were predominantly observed in participants CP5, CP9, and CP7, respectively, indicating that these clusters may be subject specific. Of the 7 TD clusters, all were present in both groups and there were no TD clusters that were subject specific.

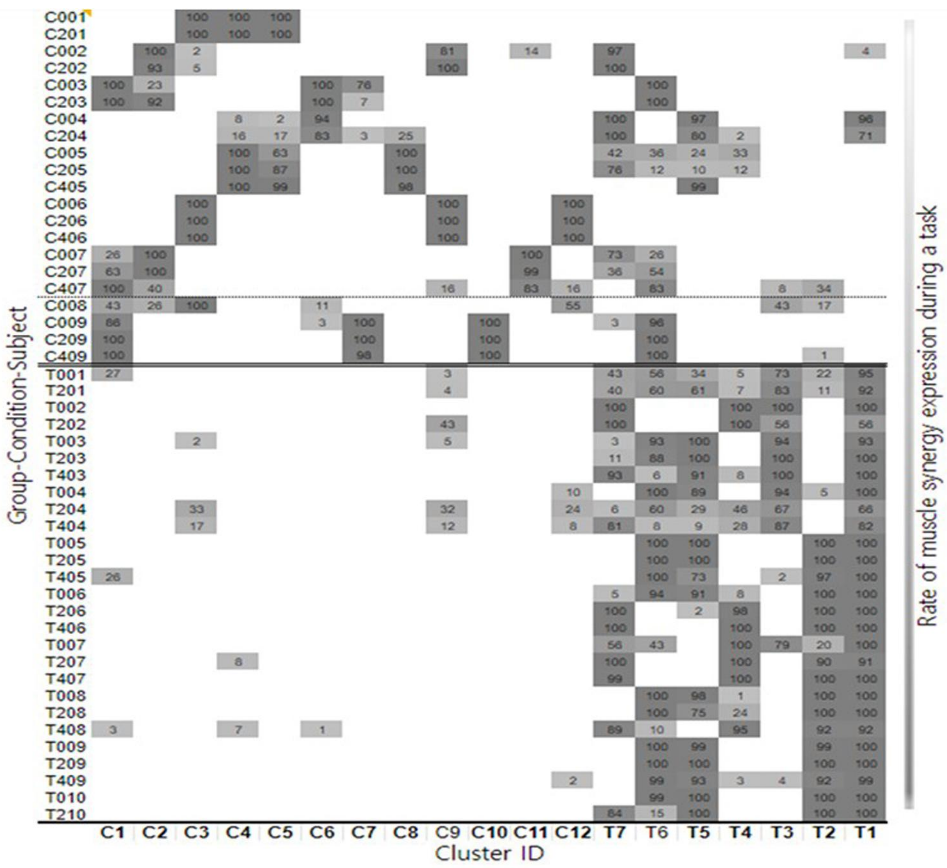

FIGURE 2
Summary for each participant and cluster showing the distribution of individual synergy structures and how these change with body weight support conditions.

3.2.2 Muscle synergy structure changes with BWS

With increasing BWS, 21.9% of synergies showed changes in TD and 11.5% in CP with most reflecting reduced activation in the Tibialis Anterior and Extensor Hallucis Longus muscles during weight acceptance. We then quantified changes in synergies within clusters without and with BWS for each participant by calculating a weight-averaged z score for each condition (29). Similar to the cluster z score, a greater positive shift in the individual weight-averaged z score with an increase in BWS indicated that a participant's synergies were becoming more CP-specific, and a greater negative shift signified that these were becoming more TD-specific. The mean shift from 0% to 20% BWS was minimal but in the direction of CP for both groups, with a slight mean shift in the direction of TD from 20% to 40% for those able to do that condition. Z scores are summarized in Table 2 by group and condition. We also calculated a mean z score across all conditions. All differed between groups as expected ($p < 0.001$). Yet, there was no significant main effect for BWS conditions for the 0, 20% or for the 0, 20% and 40% BWS cohorts ($p = 0.29$ and 0.67, respectively) or interaction ($p = 0.43$ and 0.38, respectively).

3.3 Gait analysis results

3.3.1 Temporal-spatial data

There were no significant group differences at baseline (cadence $p = 0.66$, speed $p = 0.08$, dominant step distance $p = 0.61$, non-dominant step distance $p = 0.33$: $n = 8$ CP, 10 TD). There were no appreciable or significant changes in gait speed, likely because the treadmill was set at each participant's freely selected baseline speed across conditions (Table 3). However, there were some small but significant changes in cadence which decreased slightly in both groups with 20% BWS (Table 3). Cadence also decreased in both groups from 20% to 40% BWS (Table 5). For step distance, there was a slight increase on the dominant side and a larger increase in both groups on the non-dominant side with increased BWS (Table 4). Speed showed only a 0.01 m but significant decrease in the group with TD with 40% BWS.

3.3.2 GDI values

GDI values could only be calculated for baseline and 20% BWS because there was a frequent loss of markers in the 40% BWS condition. The groups had significantly different GDI values, with the group with CP tending to improve on the dominant side only while the group with TD had marginally lower scores on both sides, with interaction p values close to but not reaching significance (Table 2; Figure 3).

3.4 EEG results

We computed ERD data for alpha (mu) and beta frequencies in the non-dominant and dominant motor clusters. Not all participants had an independent component in one or both clusters (a total of 9 TD and 5 CP had cortical sources in the non-dominant motor cluster, and a total of 9 TD and 7 CP had sources in the dominant motor cluster; 8 TD and 5 CP had sources in both hemispheres). Both TD and CP showed significant desynchronization during baseline and 20% BWS walking in both hemispheres compared to standing, indicating increased cortical activity. Comparing between 20% BWS and no BWS, there were differential effects over the gait cycle between groups whereby relative desynchronization in mu band was present in the nondominant but not dominant hemisphere in TD (Figure 3). In CP, relative mu and beta desynchronization in 20% BWS was present both hemispheres and was stronger on the nondominant side (Figure 4).

TABLE 3 General linear mixed model results for the temporal spatial gait measures from the baseline and 20% BWS conditions (bold = $p < 0.05$).

Effects	Cadence	Step distance	Speed
BWS condition	**0.02**	0.26	0.33
Condition X group	0.57	0.31	0.23
Limb	–	0.44	–
Limb X group	–	0.66	–
Condition X limb	–	**0.02**	–
Group (between)	0.69	0.38	0.23-
Cond X limb X group	–	0.08	

TABLE 4 Condition, group, and limb effects for cadence, step distance, speed.

Effects	Cadence	Step distance	Speed
Condition	**0.03**	**0.054**	**0.04**
Condition X group	0.79	0.62	0.22
Limb		0.85	
Limb X group		0.92	
Condition X limb		0.33	
Condition X limb X group		0.60	
Group	0.46	0.41	0.08

Bold indicates $p < 0.05$.

TABLE 5 General linear mixed model results for the temporal spatial gait measures from the 20% to 40% BWS conditions (bold = $p < 0.05$).

Effects	Cadence
BWS condition	0.68
Condition X group	0.06
Limb	0.23
Limb X group	0.16
Condition X limb	0.27
Cond X limb X group	0.07
Group	**0.001**

3.4.1 GLM results for EEG during BWS by group

The only significant result was a triple interaction (group X hemisphere X condition) in the beta band (Table 6). This was explained by differential effects of BWS by hemisphere across groups (Table 2) wherein the largest increase in beta ERD with BWS was in the dominant hemisphere in CP and the nondominant hemisphere in TD whereas there was a slight decrease for TD and slight increase

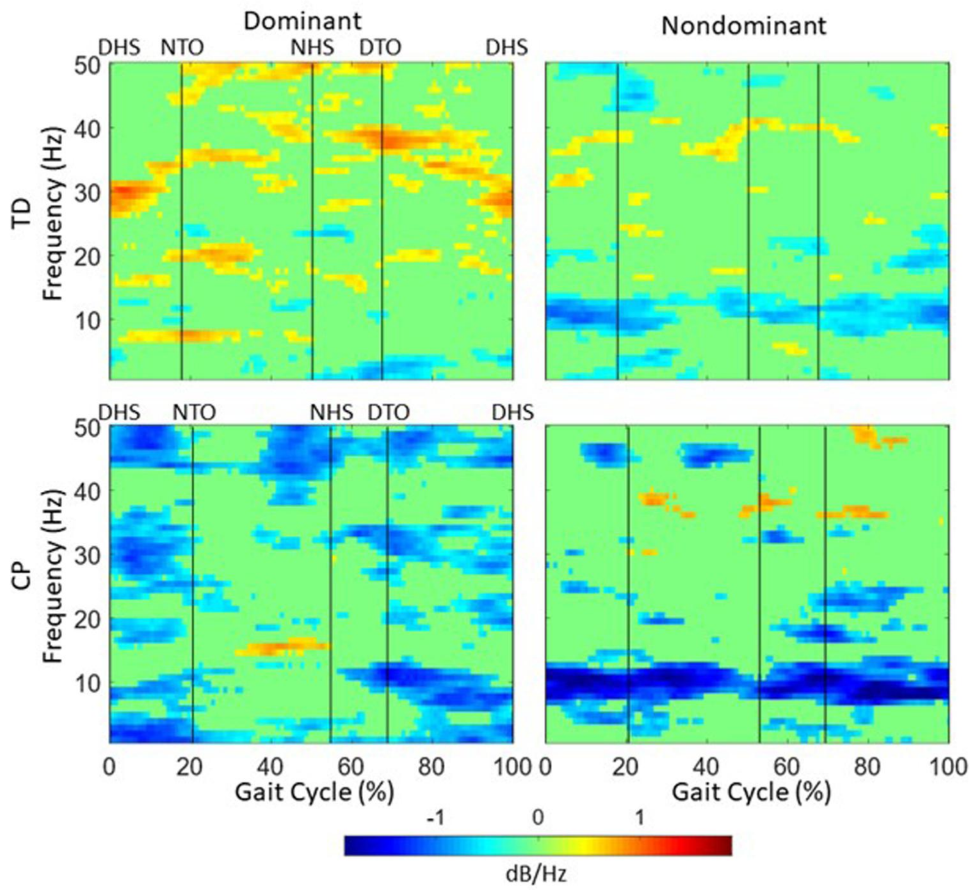

FIGURE 3
Mean GDI values for the groups with typical development (TD) or cerebral palsy (CP) for the 0% and 20% body weight support (BWS) conditions.

for CP in the dominant and nondominant hemispheres, respectively.

3.5 Correlations between neural and biomechanical measures

The baseline correlations in the entire sample showed several relationships across neural and biomechanical measures with the relationship between the synergy z score and the GDI the most notable with more TD-like scores related to more TD-like kinematics. When the same measures were correlated within groups, only the relationships between Beta ERD and gait speed remained for CP ($r = 0.84$; $p = 0.04$; Table 7).

4 Discussion

The main question addressed here was whether body weight support had a positive effect on neural and/or biomechanical measures during gait in children with unilateral CP. All participants with CP were high-functioning, able to walk independently on a treadmill with mean baseline gait speed comparable to the TD group. Neural outcome measures included muscle synergy analyses (number of synergies, walk-DMC values, and z-scores) and motor related cortical activation in bilateral sensorimotor regions during 0% and 20% BWS conditions. Biomechanical measures included temporal-spatial gait parameters and the Gait Deviation Index. Baseline neural measures were also correlated with biomechanical measures to examine interrelationships across variables.

For the sake of comparison across conditions, participants were not allowed to hold onto the treadmill rails, which affected protocol completion rates: one participant with CP could not perform any BWS condition, while four with CP and three with TD could not complete the 40% BWS condition. Interestingly, participant characteristics (age and GMFCS levels) and baseline measures (gait data and synergy numbers) did not differ between those who could and could not complete this condition.

4.1 Muscle synergy number or walk-DMC

Muscle synergy numbers and Walk-DMC are established measures of motor control complexity, with children with CP typically utilizing fewer synergies, indicating simplified control

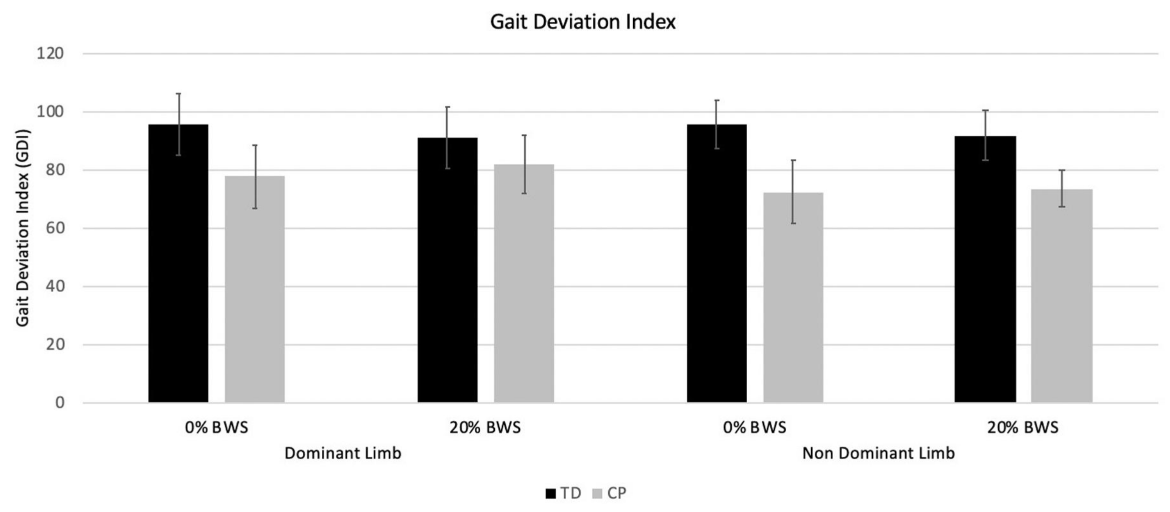

FIGURE 4
Relative ERD/ERS over the gait cycle between the 20% BWS and no BWS walking for TD (top) and CP (bottom) groups in the dominant (left) and nondominant (right) sensorimotor clusters. Negative values indicate larger ERD in 20% BWS. Plots are masked for significance ($p < 0.05$) with nonsignificant values set to 0 dB.

TABLE 6 General linear mixed model results for the EEG event-related desynchronization (ERD) measures for the alpha or mu and beta frequency bands from the 0 to 20% BWS walking conditions (bold = $p < 0.05$).

Effects	Mu ERD	Beta ERD
BWS condition	0.15	0.30
Condition X group	0.25	0.67
Hemisphere	0.53	0.51
Hemisphere X group	0.50	0.61
Condition X hemisphere	0.60	0.64
Condition X hemisphere X group	0.50	**0.048**
Group	0.44	0.18

TABLE 7 Significant correlation and associated *p*-values between baseline neural and biomechanical measures.

Neural measures	Gait speed	GDI dominant	GDI non-dominant	Dominant step distance
Z score		−0.60 (<0.01)	−0.69 (<0.01)	
Synergy number				0.47 (0.04)
Beta ERD dominant	0.58 (0.02)			

strategies (24). However, our high-functioning CP cohort here showed no significant differences in synergy numbers between groups. This finding aligns with recent research in similar functional populations (35). Additionally, while Walk-DMC scores were lower in the CP group as expected, this measure did not significantly differentiate between groups at baseline or in response to BWS. The modifiability of muscle synergies through intervention remains an open question (36). Previous intervention studies have shown inconsistent effects on synergy numbers despite improvements in kinematic outcomes (37). Our finding of unchanged synergy numbers with BWS was similar to that of Chen et al. (43); however, unlike in our study, they found differences in synergy activation patterns, recommending these as a more sensitive measure.

4.2 Synergy structures

Analysis of group-specific z scores revealed distinct muscle synergy structures between groups that remained relatively stable across BWS conditions (ranging from 11 to 13.3 in CP and from −17.4 to −14.6 in TD), confirming their distinct control strategies. This stability in muscle synergy structures suggests that temporary BWS may not substantially alter established neuromuscular control strategies, and suggests that these coordination patterns, once established, may remain relatively fixed despite transient mechanical modifications (36). This view is consistent with (38) who stated that muscle synergies are encoded prior to the onset of walking (38) and may therefore not be modifiable; However (37), which was the first study to report on changes in synergy activations in response to interventions using a similarity score based on a sample of age-matched children with TD showed a significant change in the direction of TD for those who underwent selective dorsal rhizotomy, but not for those who had orthopaedic surgery or toxin injections. Similarly, a previous study by our group demonstrated worsening synergy patterns with another environmental modification, i.e., walking on a narrower path (26) also providing a strong counter argument along with Chen et al. (43).

4.3 Gait outcomes

The basic premise for the use of BWS in CP and other neurorehabilitation populations is to improve and repetitively practice kinematic patterns, so perhaps it is no surprise that with the minimal functional impairment in our cohort with unilateral CP, the GDI values were significantly different between groups at baseline (0% BWS) and responded differentially to BWS. As presumed, the mean GDI improved in CP on both sides although the change was larger and only significant on the dominant side, whereas this tended to worsen on both sides in the group with TD. One possible explanation for the difference in the magnitude of response across legs in CP may be that the neural control in the dominant leg was more flexible and adaptable, vs. more constrained on the non-dominant side. Interestingly, the GDI regressed from 20% to 40% suggesting that 40% was not optimal for these participants. However, it may be the case, in contrast to this sample, that participants with CP who have greater weakness, spasticity, and functional impairments (i.e., GMFCS higher than Level II) would benefit from greater weight support. Cadence decreased significantly from 0% to 20% BWS in both groups by the same amount. From 20% to 40%, cadence again significantly decreased and speed increased but almost imperceptibly.

4.4 EEG outcomes

Although all EEG desynchronization values were consistently higher in the group with CP, suggesting greater cortical activation, these were not significantly different between groups at baseline. Participants in this study were a subset of those from a slightly larger group with CP who did treadmill walking with no BWS and had significantly greater EEG activation than the group with TD (25), suggesting that the finding of no significant difference here was related to the smaller sample size. EEG activation changes from 0% to 20% BWS did show a significant triple interaction between BWS X hemisphere X group in the beta band. On average, for the TD group, brain activation tended to increase in the non-dominant hemisphere and decrease in the dominant hemisphere with increasing BWS, whereas in the group with CP activation increased in both hemispheres. Increased brain activation with task difficulty often indicates greater attention or effort was needed, which in the group with TD was mostly seen to be focused on the non-dominant hemisphere which controls their non-dominant leg. In those with unilateral CP, bilateral activation is more commonly observed due to retention of ipsilateral pathways and brain reorganization post-injury (39).

4.5 Correlations among neural and biomechanical measures

When evaluating interrelationships across neural and biomechanical outcomes, *z* scores showed moderate inverse correlations with the dominant and non-dominant GDI, indicating that synergy structures more similar to TD were associated with better kinematics. Beta ERD in the dominant hemisphere was moderately correlated with gait speed in the group as a whole. This was the only significant relationship retained in the within group correlations where the group with CP showed strong correlations between Beta ERD on the dominant side and gait speed. EEG data were interestingly not correlated with synergy data, even though both represent aspects of cortical control of movement. The stronger relationship of synergy structures with kinematics and their lack of a correlation with brain activation raises the question of their direct cortical control (40). This contrasts with (27) who found that movement-related slow cortical potentials (0.5–10 Hz) were linked to synergy structures, which may differ from findings in the alpha and beta bands which we investigated here How direct the link between the brain and the muscle synergies is, however, was challenged by (41) who evaluated whether spinal motor neurons, even more closely linked to cortical output, demonstrated synergies during a complex upper limb task. The motor neuron synergies they identified better discriminated individual finger forces than muscle synergies and, in a few cases, motor neurons innervating a given muscle were active in separate synergies. This indicates that cortical mapping onto muscles is not direct but involves other even closer levels of the nervous system in dimensionality reduction.

4.6 Clinical and research implications

Since stepping is essentially reflexive in spinal-lesioned animals, electrophysiological methods have been used to examine neural pathways and reflexes in response to body weight unloading (44) with one study showing that cutaneous reflexes were enhanced with unloading (45) with potentially positive effects on motor output (46). It is theorized that BWS during treadmill training supplies the injured nervous system with necessary and appropriate sensory input signals for stimulating intact spinal cord networks, most notably central pattern generators (CPG), which can be directed towards improving the control of walking (11, 47). BWS treadmill training aimed to utilize spinal reflexes and networks to facilitate stepping originally for patients with complete spinal cord injuries. Cortical involvement during walking based on task-specific EEG studies in healthy adults has been shown to be greater than previously assumed and greater engagement of brain activity during motor training may enhance effectiveness (31, 42), Thus, gait rehabilitation is more complex in disorders such as stroke and CP where cortical input is present but may be abnormal and reciprocal stepping is present but muscle activation patterns and joint motions are not as efficient, smooth or forceful as seen in those without these disorders. Weakness and poor postural control may further limit gait rehabilitation in CP and stroke but are accommodated rather than remediated by harnessed or device-provided weight support; and therefore, often need to be addressed separately.

As anticipated, the effects of BWS on muscle, brain, and gait measures in those with TD were negligible and even somewhat negative in some cases (e.g., in their GDI values which in contrast improved in CP). Gait in those with typical development is optimized for efficiency and low energy expenditure and perturbations such as reducing body weight are unlikely and unnecessary to improve mobility function. Increases in brain activation with increased BWS was also only seen in CP indicating that this required greater attention or effort in those with mobility challenges, with a negligible effect in TD likely due to the greater flexibility and adaptability of their motor control system. This study demonstrated that providing some level of BWS produces greater biomechanical similarity to gait in those with TD and thus may offer some training benefits that should be combined with strengthening and exercises to enhance postural control. However, the relationship between the level of BWS and improved clinical, neural and/or biomechanical outcomes in CP may not be entirely linear as it appears when going from 0% to 20% BWS and instead may be more curved or even an inverted-U shape as shown by improvements plateauing or even reversing in some cases with 40% BWS. From a clinical standpoint, it seems important to advise that the level be determined carefully using objective parameters when available so as to maximize the training benefit for each individual. Perhaps other more invasive interventions such as orthopaedic and neurosurgery or botulinum toxin injections that also address biomechanical and/or neural impairments common in CP will be needed for larger effects. These interventions must ultimately demonstrate that they significantly improve function or participation in children with CP to justify their use, either alone or in combination, in clinical practice.

4.7 Limitations

The major limitation here was the small sample size of those who could perform the baseline condition (0% BWS) and one or both BWS conditions (20% and 40% BWS). The study design also required that those with CP have a high level of mobility which limited the ability to detect group differences when compared to the TD group, given the similar functional walking abilities of both groups. However, several significant immediate effects of BWS on neural and biomechanical measures were identified that in some cases were similar and other cases divergent across groups. Finally, this was not a training study, so longer-term effects of BWS on these outcomes were not addressed here.

4.8 Conclusion

BWS can alter joint kinematics in CP in the direction of greater similarity to those without CP which were also related to synergy z scores, likely because the timing and magnitude of muscle activation produces joint motion patterns. These findings suggest potential clinical benefits from BWS for those with CP. EEG did relate to gait speed as a measure of function but did not vary with muscle synergies that many propose are encoded or controlled by the CNS, thus continuing the debate on the role of muscle synergies in the neural control of movement (40).

Author's note

The contributions of the NIH author(s) are considered Works of the United States Government. The findings and conclusions presented in this paper are those of the author(s) and do not necessarily reflect the views of the NIH or the U.S. Department of Health and Human Services.

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 humans were approved by National Institutes of Health - Institutional Review Board. Written informed consent for participation in this study was provided by the participants' legal guardians/next of kin.

Author contributions

RB: Formal analysis, Data curation, Methodology, Investigation, Writing – original draft. YK: Formal analysis, Data curation, Project administration, Visualization, Methodology, Investigation, Software, Writing – review & editing, Conceptualization. TB: Data curation, Project administration, Conceptualization, Methodology, Writing – review & editing, Supervision, Investigation, Software, Formal analysis. DD: Formal analysis, Data curation, Project administration, Writing – original draft, Methodology, Investigation, Supervision, Conceptualization.

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Conflict of interest

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SUPPLEMENTARY FIGURE S1
Muscle synergy activation clusters with the timing of the cluster's muscle activation across the gait cycle shown in the top left corner of each figure, peak activation time in the gait cycle (top right) and structures (bottom). The left column consists of clusters specialized for cerebral palsy, and the right one is for typical development. Subfigures are arranged based on peak activation time from top to bottom for each column. Data are expressed as mean and standard deviation. The labels of bar plots are combined with leg side and muscle names as follows: D, dominant leg; N, non-dominant leg; TA, tibialis anterior; EH, extensor hallucis longus; LG, lateral gastrocnemius; SO, soleus; RF, rectus femoris; VL, vastus lateralis; ST, semitendinosus; BF, biceps femoris.

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