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Research article

Muscle mechanics and energetics in chronic ankle instability and copers during landing: Strategies for adaptive adjustments in locomotion pattern

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A R T I C L E I N F O

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A B S T R A C T

Individuals with chronic ankle instability (CAI) and copers typically exhibited aberrant landing kinematics. Altered kinematics might lead to changes in muscle loading, potentially affecting the energy demand of locomotion. Understanding alterations in muscle mechanics and energetics during landing could enhance the rehabilitation program design. Therefore, the objective of this study was to explore the muscle mechanics and energetics of individuals with CAI, copers, and healthy controls during single leg jump landing. Three groups, CAI, copers, and healthy individuals (total $n = 66$), performed the landing task, and data on 3D motion capture, ground reaction force (GRF), and muscle activation were simultaneously collected. A musculoskeletal model was applied to estimate muscle force and mechanical power. Compared to healthy groups, individuals with CAI showed increased peak muscle forces in the gluteus maximus ($p < 0.001$), gluteus medius ($p < 0.001$), vastus lateralis ($p < 0.001$), and peroneus longus ($p < 0.001$) during landing. Whereas copers exhibited higher peak muscle forces in the vastus lateralis ($p < 0.05$), medial gastrocnemius ($p < 0.05$), soleus ($p < 0.05$), and peroneus longus ($p < 0.001$). Additionally, negative mechanical power redistribution in CAI shifted from the ankle to the hip ($p < 0.001$), while copers exhibited a similar redistribution from the ankle to the knee ($p < 0.05$). This study suggested that both CAI and copers exhibit biomechanical modifications in proximal joints. Copers showed a novel landing strategy aimed for enhancing landing stability, but with the risk of ACL injury. The identified energy redistribution observed in both CAI and copers could potentially contribute to the recurrent ankle sprains. This research facilitates a better understanding of how muscle mechanics and energy demands influence the landing pattern in individuals with CAI and copers.

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1. Introduction

Lateral ankle sprain (LAS) is a prevalent clinical condition, representing 25 % of all musculoskeletal injuries [1,2]. Despite its high incidence and its propensity to lead to various consequences during sports activities, there is a misconception that ankle sprains can heal without professional medical care. Consequently, up to 80 % of individuals initially experiencing ankle sprains subsequently encounter secondary sprains, with approximately 40 % progressing to chronic ankle instability (CAI) [3]. CAI is characterized by recurring ankle sprains, a high frequency of the ankle giving way, and biomechanical alterations [4,5]. Jay et al. introduced a revised model of CAI, proposing three categories of impairments: pathomechanical impairments, sensory-perceptual impairments, and motor-behavioral impairments [6]. The interplay of these impairments results in enduring ankle instability, predisposing individuals to recurrent sprains.

Copers are individuals who, after experiencing an initial ankle sprain, successfully recover and maintain joint function. They are typically able to effectively manage the potential instability and symptoms that may follow the injury by using rehabilitation, appropriate movement control, and adaptive changes to reduce the risk of subsequent injuries [6]. In recent years, biomechanical researchers have investigated potential deficits in locomotion patterns during dynamic tasks among copers, healthy individuals and those with CAI [7]. Researchers have reported that increased hip flexion angles and hip joint stiffness in individuals with CAI during a drop landing task [8]. It has been established that the utilization of the 'hip strategy' is integral to their daily activities in individuals with CAI. This strategy helps maintain sagittal and frontal plane stability during daily activities, effectively resisting external perturbations and preventing injuries. Another study investigated locomotive biomechanics during walking in individuals with CAI, copers, and healthy participants [9,10]. This study revealed that CAI individuals exhibited greater hip flexion at initial contact, increased knee flexion and ankle inversion, but less knee flexion moment at toe-off compared to copers. In contrast, copers had greater knee flexion and ankle inversion but less ankle plantarflexion at initial contact compared to healthy individuals. The increased ankle inversion observed in individuals with CAI may result from adaptations in the central nervous system following ankle sprains, giving rise to associated neuromuscular deficits [11].

Aberrant kinematics have been clearly demonstrated in both CAI and coper participants [7]. These abnormal kinematic characteristics in CAI and copers may lead to changes in muscle loading driven by neuromuscular coordination, potentially presenting greater challenges to muscle function [12,13]. Alterations in muscle function could impact the locomotion patterns of CAI and copers, an area that remains unexplored. Musculoskeletal modeling is a valuable tool for studying muscle roles during dynamic tasks especially for subclinical and young ankle sprain patients, providing a non-invasive alternative [14]. Recently, a simulation compared muscle force differences between CAI and healthy groups during landing [15]. It was reported that the compensatory strategy of CAI is reflected in force production in proximal muscles, but more information about copers remains unclear. Understanding muscle dysfunctions in copers is crucial for preventing secondary injuries and designing effective rehabilitation programs. Moreover, it assists in interpreting potential adaptive movement patterns observed in individuals identified as copers.

Ideally, muscles generate power during movement based on the force-length-velocity relationship [16]. Abnormal kinematic characteristics result in changes to the contraction velocity and length requirements of muscles, thus affecting muscle power, as well as influencing energy dissipation and redistribution processes [17]. Few studies have explored potential differences in movement patterns related to energy distribution strategies between individuals with CAI and copers during landing. Kim et al. examined difference in landing and cutting strategies among the CAI, copers, and healthy groups in terms of joint stiffness and joint power [18,19]. However, for researching the energy distribution of CAI, copers during landing, the simultaneous execution of landing and sidecut tasks may introduce confounding factors. Therefore, these studies provide limited information regarding lower extremity energy dissipation strategies for CAI, copers, and healthy individuals during landing, necessitating further detailed clarification.

The primary goal of rehabilitating patients with ankle sprains is to facilitate the recovery of motor-perceptual functions by promoting central nervous system reorganization and compensation [20]. However, traditional ankle rehabilitation training falls short in terms of frequency and intensity, primarily due to time and resource constraints. This limitation contributes to less favorable patient outcomes and places an increased burden on ankle trainers [21]. The development of an ankle rehabilitation robot could improve the effectiveness of ankle sprain rehabilitation by offering precise control over training methods and adjusting complex force parameters [22]. Identifying the muscle mechanics and energy dissipation or redistribution characteristics in the lower extremities of individuals with CAI and copers could potentially contribute to the development of improved exoskeletons or assisted rehabilitation devices to restore a more natural loading environment during patients' daily activities.

Therefore, this study aimed to investigate muscle mechanics and energetics characteristics during single leg jump landing in individuals with CAI, copers, and healthy individuals. Based on previous studies, it is reasonable to hypothesize that copers, like CAI, may employ compensatory patterns to make slight adjustments to their movement patterns. But CAI and copers may exhibit different muscle mechanics and energetics characteristics during a landing task.

2. Methods

2.1. Participants

Utilizing G*power software for a priori power analysis [23], with statistical power and significance level fixed at 0.80 and 0.05, respectively, the results revealed that a minimum sample size of 66 subjects was necessary to meet the criteria for a medium effect size of 0.25. This effect size was chosen based on existing literature, where similar studies comparing individuals with chronic ankle instability (CAI) and copers often report medium effect sizes in this range. We provided participants within the academic institution with the self-administered Cumberland Ankle Instability Tool (CAIT) [24]. Each participant completed a questionnaire and documented anthropometric parameters, the frequency of lateral ankle sprains, and the history of medical consultations (Table 1). Finally, we recruited 66 male participants, including those with CAI, copers, and healthy controls, with 22 individuals per group, based on their scores.

According to the consensus statement from the Inter-national Ankle Consortium [25], CAI was defined as individuals has 1) CAIT score≤24. 2) A record of ankle sprains resulting from trauma necessitating two or more medical consultations. 3) Reported experiencing recurrent lateral ankle sprains lasting at least six months or expressed concerns regarding potential dysfunction of the ankle. Copers were characterized as individuals who had experienced an initial sprain in the past but did not report any complaints of instability [6]. Those people also have 1) CAIT score had to range between 25 and 28. 2) no episodes of re-injury. 3) The initial ankle sprain should have taken place a minimum of 12 months before enrollment in the study. 4) Caused the interruption of planned physical activity for a minimum duration of one day. Moreover, individuals with no history of lateral ankle sprains and a CAIT score of ≥29 were included in the control group. Exclusion criteria encompassed a history of prior lower extremity surgical interventions, lower extremity fractures, bilateral ankle sprains, or failure to meet the inclusion criteria for the CAI, copers, and control groups. To mitigate the influence of limb dominance on the experiment, all participants included in the study had their right limb designated as the dominant limb, including the injured limb. The dominant limb was identified as the preferred leg for kick a ball [26].

All participants were apprised of the study's purpose, requirements, and procedures, and each participant provided written informed consent. The study protocol received approval from the Ningbo University Scientific Research Ethics Committee (Approval Number: RAGH20220620).

2.2. Data collection procedures

Each participant was equipped with 39 reflective skin markers for motion tracking, as illustrated in Fig. 1A. Following SENIAM guidelines to place the EMG sensors [27], we prepared the acquisition area by scraping the hair near the skin and cleaning it with alcohol to reduce skin-electrode interface impedance. Five Electromyographic (EMG) sensors (Delsys, Boston, MA, USA) were used to quantify the muscle activations by positioning over the muscle bellies of the soleus, medial gastrocnemius, lateral gastrocnemius, tibialis anterior, and peroneus longus (Fig. 1B).

Before commencing the formal experiment, a 15-min window was provided for all participants to complete a dynamic warm-up and practice single leg jumping landing to familiarize themselves with the task. They were allowed to practice without any feedback or instructions on landing technique to avoid influencing their natural landing patterns. During the task session, participants were directed to maintain a stance on their injured leg, placing themselves at a distance roughly equivalent to one leg's length (measured along the line from the greater trochanter to the lateral malleolus) from the force plate. They were additionally guided to position their hands on both sides of their hips, aiming to reduce the impact of arm swing on the experimental procedure. Following that, they were required to leap over a hurdle with a height of 15 cm, landing on the same leg. They were then required to promptly stabilize and maintain balance for a duration of 5 s, while maintaining a forward gaze (Fig. 1C).

All tasks were performed barefoot, and each participant was tasked with accomplishing three effective trials of single leg jump landings. A trial was considered successful if participants landed on the force plate with their entire foot. In instances where participants struggled to sustain balance or introduced additional movements during the single leg jump landing, that specific trial was categorized as unsuccessful. The experimental procedure persisted until a cumulative total of three successful trials were obtained, with 3-min intervals between each trial.

The Vicon motion capture device with eight infrared cameras (Vicon Metrics Ltd., UK) was used to collect three-dimensional positions of the reflective markers at a 200 Hz sampling frequency. Simultaneously, the ground reaction force (GRF) data were collected using a 1000 Hz AMTI force plate (AMTI, Watertown, UAS), and muscle activation data were recorded with EMG sensors at a 1000Hz sampling frequency.

Table 1
Participant demographics.

	CAI (n = 22) Mean (SD)	Copers (n = 22) Mean (SD)	Healthy (n = 22) Mean (SD)	*P*-value
Age(year)	23.6(1.2)	23.5(1.4)	22.8(1.4)	0.349
Mass(kg)	80.6(5.6)	79.5(6.5)	80.7(6.1)	0.864
Height(cm)	179.8(4.5)	180.2(4.1)	178.7(4.0)	0.643
CAIT(score)	20.8(2.6)	26.3(1.1)	29.6(0.5)	<0.001
Ankle sprains(times)	3.3(1.1)	1	0	<0.001
Leg length (cm)	94.5(3.1)	94.6(3.7)	93.7(4.6)	0.794
Landing time(s)	0.24(0.07)	0.22(0.04)	0.20(0.07)	0.102

Note: SD: standard deviation; CAI: chronic ankle instability; CAIT: Cumberland Ankle Instability Tool.

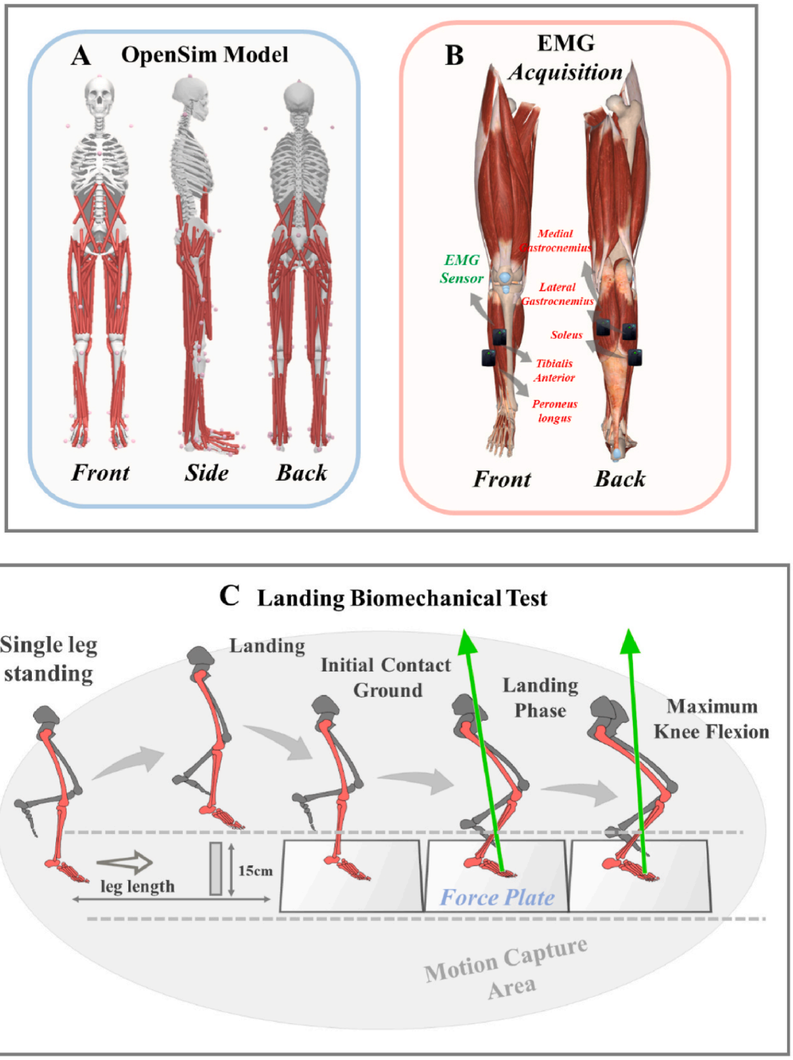

Fig. 1. Presentation of the musculoskeletal model and a comprehensive outline of the experimental procedures employed in the study. **(A)** Depiction of the position of reflective marker on the constructed musculoskeletal model. **(B)** Visualization of the placement of the EMG tests on the lower limbs of human subjects. **(C)** Illustration of the process of single leg jump landing biomechanics test.

2.3. Data processing procedures

Three-dimensional marker trajectories and ground reaction force data were identified and captured using Vicon Nexus 2.14.0 software, and subsequently exported in a c3d format file (Fig. 2B). Following Winter's research on the selected filter frequency [28], we determined the optimal signal-to-noise ratio through residual analysis on data subsets. Subsequently, we implemented fourth-order zero lag Butterworth low-pass filters with cutoff frequencies set at 12 and 20 Hz for filtering both kinematic and kinetic data. The initial step involved subjecting the raw surface EMG signal to band-pass filtering using a fourth-order Butterworth filter within the 10–400 Hz frequency range. Following this, full-wave rectification was applied, succeeded by low-pass filtering with a cut-off frequency of 6 Hz [29]. Simultaneously, the EMG amplitude was normalized by maximum voluntary contraction (MVC) to obtain each muscle's activation level.

2.4. Musculoskeletal modeling

Biomechanical parameters throughout this research were computed using OpenSim software (Stanford University, Stanford, CA, USA) [14]. A musculoskeletal model with 23 degrees of freedom and 92 musculotendon actuators was used to perform all musculoskeletal simulations [30]. Subject-specific musculoskeletal models were generated using the scale tool in OpenSim, involving the scaling of a generic musculoskeletal model (Fig. 2C). Additionally, we updated the maximum isometric force of the musculotendon actuators by estimating subject-specific muscle characteristic parameters. **(**Eqs. (1), (2)**)**.

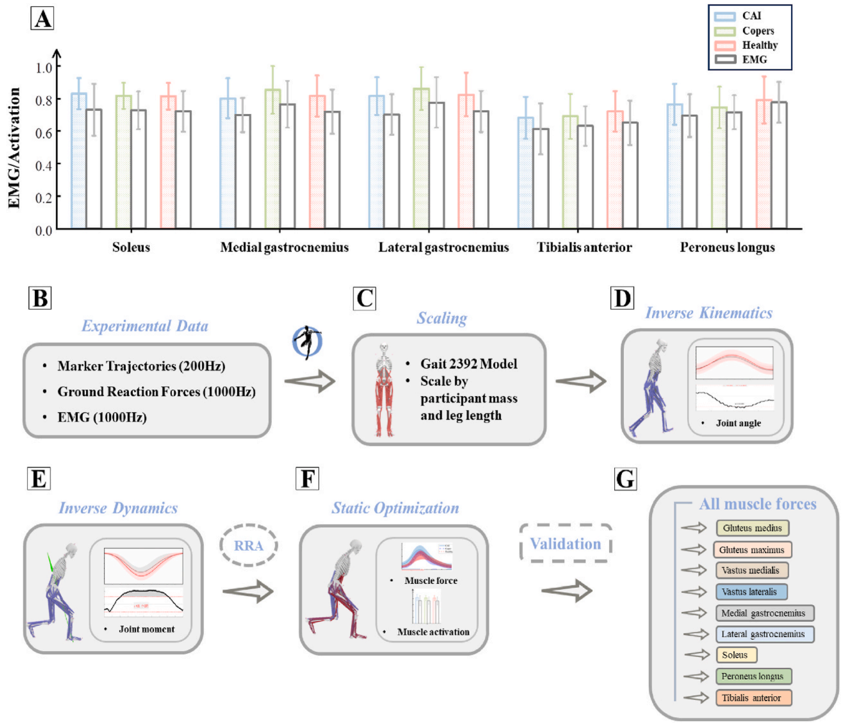

Fig. 2. The procedural flow of the musculoskeletal modeling process employed for muscle force calculation. **(A)** Verification through a comparison of the measured muscle activation patterns with the simulated activations. **(B)** Importing experimental data into Opensim software. **(C)** Scaling of model. **(D)** Inverse Kinematics. **(E)** Inverse Dynamics. **(F)** Static Optimization. **(G)** Calculate all muscle forces. RRA: Residual Reduction Algorithm.

$$V_{total} = 47mh + 1285 \quad (1)$$

$$F_o^m = \sigma_o^m \frac{\varnothing^m V_{total}}{l_o^m} \quad (2)$$

The total muscle volume of the lower limb (V_{total}) was initially estimated using the regression equation (Eq. (1)), which was based on the height (h) and weight (m) of each subject. Individual muscle volume fractions ($\varnothing^m$) and muscle lengths were determined through Geoffrey et al. studies [31]. The optimal muscle fiber length $\left(l_o^m\right)$ for each muscle was then calculated based on the ratio of muscle fiber length to muscle length [32]. The tension of each muscle was set to 60 N/cm^2 [14]. Finally, the maximum isometric muscle force (F_o^m) of the musculotendon actuator was updated according to Eq. (2). Following the best practice guidelines proposed by Hicks et al. [33], the efficacy and precision of the musculoskeletal model were verified by comparing the muscle activation outcomes obtained from the EMG sensor with those derived from musculoskeletal modeling simulations (Fig. 2A). In cases where discrepancies were observed between the two data sources, priority was given to the EMG data due to its direct reflection of muscle activation. But when EMG data were noisy or unreliable, we relied on the musculoskeletal model results to maintain data consistency. Finally, no significant differences were observed between the muscle activation data from EMG and musculoskeletal modeling simulation.

The inverse kinematic tool was employed for joint angle calculation, utilizing a weighted least-squares optimization approach aimed at minimizing the disparity between the model and experimental marker positions (Fig. 2D). Joint moment for each degree of freedom of the model was calculated using inverse dynamics tools (Fig. 2E). To enhance the accuracy of our model by minimizing dynamic inconsistency, we employed a residual reduction algorithm. The calculation of muscle forces utilized a static optimization algorithm, decomposing joint moments into individual muscle forces by minimizing the sum of squared muscle activations (Fig. 2F). Ultimately, we computed the nine muscle force variables of interest to reflect muscle loading (Fig. 2G). The schematic representation of the workflow for the musculoskeletal modeling process is illustrated in Fig. 2. Joint power at each time point was computed as the product of angular velocities and joint moment [34]. To calculate the average mechanical work values for each joint, we integrated the negative portion of the joint power curve [35]. The average mechanical power for each joint was derived by dividing the average mechanical work values by the average landing time for the trial [36]. To calculate the total limb's average mechanical power, we summed the average mechanical power values at each joint. These values were then used to determine the percentage contribution of each individual joint to the total limb's average mechanical power.

2.5. Outcome variables

The outcome variables of interest included the muscle force and mechanical power of various groups (CAI, copers, and healthy). All biomechanical variables in this study were meticulously computed and analyzed within the defined landing phase. This phase, characterized by the duration from the moment of initial contact to the point of maximum knee flexion angle, commenced at the time when the vertical ground reaction force (VGRF) surpassed 20 N [37]. For time-dependent variables, the data were normalized to 101 points, corresponding to the 0–100 % range of the landing phase, facilitated by a customized MATLAB code. All the force-related parameters (i.e., muscle force) were normalized to the body weight, while other parameters (i.e., mechanical work, mechanical power, and joint moment) were normalized to body mass. This normalization procedure was consistently applied across all groups to account for potential differences in body weight, ensuring that any discrepancies between groups did not affect the comparability of the data.

2.6. Statistical analysis

Normality and homogeneity of variances in the outcome variables were evaluated through the Shapiro-Wilk and Levene's tests, respectively. Upon confirmation of the assumptions of data normality and homogeneity of variances, between-group differences were examined via a one-way analysis of variance (ANOVA). In cases where the assumption of data normality was not met, non-parametric Kruskal-Wallis H-tests were employed for the analysis of dependent variables. Subsequently, post-hoc multiple comparisons with Bonferroni correction were carried out to investigate pairwise differences. The significance level was set at 0.05. We utilized eta-squared (η^2) effect sizes to quantify the alterations in the outcome variables across various groups (Eq. (3)). Effect size values were interpreted based on established criteria: values within the range of 0.04–0.25 denoted a small effect, those from 0.25 to 0.64 indicated a medium effect, and values exceeding 0.64 signified a large effect.

$$\eta^2 = \frac{SS_b}{SS_t} \tag{3}$$

For a more comprehensive insight into the temporal changes in muscle force during the landing phase, we conducted Statistical Parametric Mapping (SPM) analysis to compare muscle forces at different time points [38]. The SPM also involved the computation of the critical threshold using a random field theory, along with the generation of SPM{T} curves. A surpassing of the critical threshold by SPM{T} indicated a significant difference in muscle force between groups at that specific time point. MATLAB (MathWorks, MA, USA)

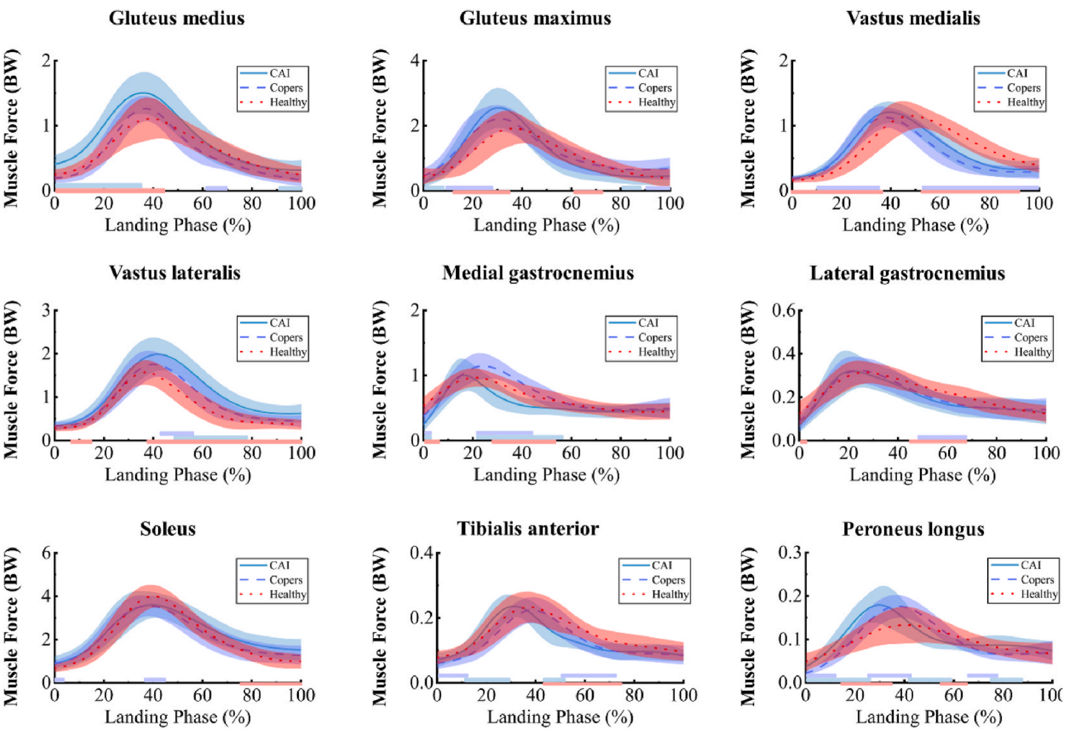

Fig. 3. Mean ± SD normalized time-series muscle forces in people with chronic ankle instability (CAI), copers and healthy controls during single leg jump landing. The blue line illustrates the outcome of the SPM analysis comparing CAIs and copers, showing statistical difference during the corresponding landing phase. Likewise, the red line illustrates the SPM analysis findings between the CAI and healthy groups and the purple line illustrates the SPM analysis results between the copers and healthy groups.

was employed for all statistical analyses.

3. Results

3.1. Participant demographics

No significant differences were observed among groups with respect to age, height, weight, leg length, and landing time. However, CAIT score in the CAI group was significantly lower than copers and healthy groups, while the number of ankle sprains in the CAI group was significantly higher than copers and healthy groups (Table 1).

3.2. Muscle force

In the SPM1d analysis results, ANOVA indicated statistically significant differences between all muscle groups (Fig. 3). Specifically, post-hoc analysis demonstrated that, for the gluteus medius, CAI was significantly higher than copers during 0%–38 % ($p < 0.001$) and 91%–100 % ($p < 0.05$) of the landing phase. Furthermore, CAI was significantly higher during 0%–44 % ($p < 0.001$) phase when compared to the healthy individuals. Conversely, Copers was significantly lower than healthy during 61%–70 % ($p < 0.05$) of the landing phase. For the gluteus maximus, CAI was significantly lower than copers during 0%–4% ($p < 0.05$) and 80%–88 % ($p < 0.001$) of the landing phase. additionally, CAI was significantly lower compared to the healthy during 61%–72 % ($p < 0.001$) of the landing phase, but significantly higher during 13%–35 % ($p < 0.001$). Conversely, copers were higher than healthy during 9%–28 % ($p < 0.001$) and 90%–100 % ($p < 0.05$) of the landing phase. For the vastus medialis, both CAI and copers profiles were similar, with CAI was significantly higher during 0%–38 % ($p < 0.001$) of the landing phase and significantly lower during 52%–92 % ($p < 0.001$) phase compared to the healthy. However, copers were significantly higher during 10%–36 % ($p < 0.001$) of the landing phase and significantly lower during 52%–100 % ($p < 0.001$) phase compared to the healthy.

For the vastus lateralis, CAI was significantly higher than copers during 49%–79 % ($p < 0.001$) of the landing phase, and significantly higher than healthy during 38%–100 % ($p < 0.001$) and 7%–15 % ($p < 0.05$) of the landing phase. In contrast, copers were significantly higher than healthy during 42%–55 % ($p < 0.001$) of the landing phase. For the medial gastrocnemius, CAI was significantly lower than copers during 0%–2% ($p < 0.05$) and 21%–57 % ($p < 0.001$) of the landing phase. Additionally, CAI was significantly lower compared to the healthy during 0%–6% ($p < 0.05$) and 25%–52 % ($p < 0.001$) of the landing phase. On the other hand, copers were significantly lower during 0%–2% ($p < 0.05$) but significantly higher during 21%–43 % ($p < 0.001$) of the landing phase compared to the healthy. For the lateral gastrocnemius, CAI and copers had similar force profiles, but CAI was significantly lower than healthy during 0%–2% ($p < 0.05$) and 45%–68 % ($p < 0.001$) of the landing phase, while copers were also significantly lower than healthy during 49%–68 % ($p < 0.001$).

For the soleus, CAI and copers profiles were similar. Compared to the healthy, CAI was significantly higher during 75%–100 % ($p < 0.001$). Conversely, copers were significantly higher than healthy during 0%–3% ($p < 0.05$) but significantly lower during 38%–46 % ($p < 0.05$). For the tibialis anterior, CAI was significantly higher than copers during 11%–29 % ($p < 0.001$) of the landing phase but significantly lower during 44%–52 % ($p < 0.05$). When compared to the healthy, CAI was significantly lower during 42%–75 % ($p < 0.001$) of the landing phase. Conversely, copers were significantly lower during 0%–12 % ($p < 0.05$) and 50%–72 % ($p < 0.001$) of the landing phase. For the peroneus longus, CAI was significantly higher than copers during 0%–26 % $p < 0.001$) and 76%–88 % ($p < 0.001$) of the landing phases, while it was significantly lower than copers during 42%–58 % ($p < 0.001$) of the landing phases. CAI was also significantly higher than healthy during 15%–35 % ($p < 0.001$) of the landing phases but significantly lower during 54%–65 % ($p < 0.05$). On the other hand, copers were significantly lower than healthy during 0%–13 % ($p < 0.05$) and 66%–77 % ($p < 0.001$) of the landing phases but significantly higher at 25%–43 % ($p < 0.001$) in the landing phase.

No statistically significant differences were found in the peak muscle force of vastus medialis ($p = 0.314$, $\eta^2 = 0.024$), Lateral gastrocnemius ($p = 0.317$, $\eta^2 = 0.023$), and tibialis anterior ($p = 0.710$, $\eta^2 = 0.007$) among the CAI, copers, and healthy (Table 2, Fig. 4). However, for the gluteus medius, individuals with CAI had significantly higher peak muscle force during landing phase

Table 2
Detailed results of the peak muscle forces between those with CAI, copers and healthy controls during single leg jump landing.

Muscle force parameters (BW)	Group			*F*	*P*-value	*ES*
	CAI Mean (SD)	Copers Mean (SD)	Healthy Mean (SD)			
Gluteus medius	1.57 ± 0.33	1.31 ± 0.21	1.21 ± 0.26	14.681	<0.001	0.534
Gluteus maximus	2.67 ± 0.58	2.37 ± 0.36	2.17 ± 0.43	9.482	0.001	0.164
Vastus medialis	1.25 ± 0.16	1.18 ± 0.17	1.22 ± 0.19	1.173	0.314	0.024
Vastus lateralis	2.07 ± 0.39	1.87 ± 0.25	1.60 ± 0.28	21.424	<0.001	0.265
Medial gastrocnemius	1.07 ± 0.20	1.18 ± 0.19	0.99 ± 0.14	8.645	<0.001	0.458
Lateral gastrocnemius	0.36 ± 0.08	0.35 ± 0.06	0.34 ± 0.05	1.164	0.317	0.023
Soleus	3.92 ± 0.62	3.77 ± 0.53	4.12 ± 0.51	3.175	<0.05	0.064
Peroneus longus	0.19 ± 0.04	0.18 ± 0.02	0.15 ± 0.04	15.468	<0.001	0.258
Tibialis anterior	0.25 ± 0.03	0.24 ± 0.03	0.26 ± 0.04	2.599	0.080	0.007

Note: SD: standard deviation; ES: effect size; BW: body weight; CAI: chronic ankle instability.

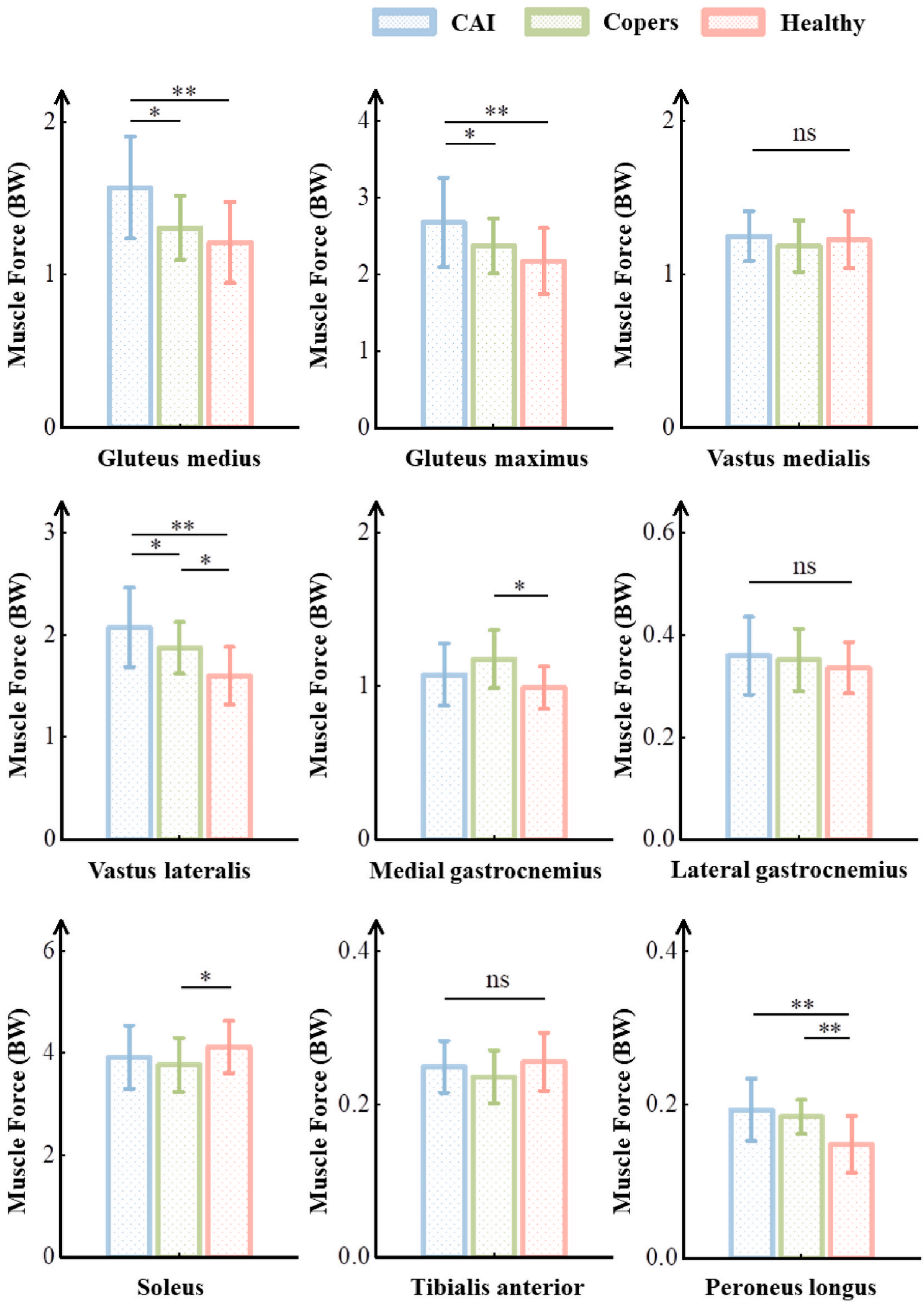

Fig. 4. Mean ± SD peak muscle forces in people with CAI, copers and healthy controls during single leg jump landing. ns: no significant. *: significant difference with $p < 0.05$. **: significant difference with $p < 0.001$.

compared to both copers and healthy individuals ($p < 0.05$, $p < 0.001$, respectively). Similarly, for the gluteus maximus, the peak muscle force was significantly higher during the landing phase in CAI compared to the healthy ($p < 0.001$), while copers showed a significantly higher peak muscle force as well ($p < 0.05$) compared to the healthy. For the vastus lateralis, CAI had a significantly higher peak muscle force when compared to copers and healthy ($p < 0.05$, $p < 0.001$, respectively). Likewise, copers had a significantly higher peak muscle force ($p < 0.001$) compared to the healthy. For the medial gastrocnemius, copers had a significantly higher peak muscle force ($p < 0.05$) than the healthy during the landing phase. In contrast, for the soleus, the healthy showed a significantly higher peak muscle force ($p < 0.05$) than copers. It's worth noting that no significant differences were observed between CAI and copers or healthy. For the peroneus longus, both CAI and copers exhibited significantly higher peak muscle force compared to the healthy ($p < 0.001$, $p < 0.001$, respectively).

3.3. Mechanical power

For the average mechanical power (Table 3, Fig. 5), individuals with CAI exhibited a significantly higher ($p < 0.001$, $p < 0.001$, respectively) average hip mechanical power compared to both copers and healthy individuals, leading to a significant increase in total limb average mechanical power. Specifically, for the ankle average mechanical power, no significant differences were found between CAI and copers, while individuals in the healthy were significantly higher than CAI and copers during landing phase ($p < 0.001$, $p < 0.001$, respectively). In contrast, for the knee average mechanical power, a significant difference was observed only between copers and healthy, with copers were significantly higher ($p < 0.05$) than healthy during the landing phase.

Regarding the relative contribution of each joint to the total limb average mechanical power (Table 3), across all groups, the knee exhibited the highest contribution, with the ankle joint showing a significantly lower ($p < 0.001$) relative contribution in CAI compared to the healthy but a significantly higher contribution than healthy and copers in the hip joint ($p < 0.001$, $p < 0.001$, respectively). Additionally, the relative contribution of the knee joint was significantly higher in copers compared to both CAI and healthy ($p < 0.001$, $p < 0.05$, respectively), while the relative contribution of the ankle joint was significantly lower ($p < 0.001$) than the healthy.

4. Discussion

This study examined muscle mechanics and energetics in individuals with chronic ankle instability (CAI), copers and healthy controls during single leg jump landing. To the best of our knowledge, it is the first study to comprehensively quantify muscle force and joint mechanical power during single leg jump landing in individuals with CAI, copers and healthy participants. Our result revealed that: (1) Both CAI and copers exhibit altered muscle mechanics and energetics at the proximal joints. (2) From the perspectives of muscle mechanics and energetics, copers might exhibit a predominant 'knee strategy' during landing, attributed to a compensatory strategy in the extensor and flexor muscles. (3) The significant difference in peak muscle force observed in the soleus and the peroneus longus might constitute the primary factor contributing to the differences in landing patterns between individuals with CAI and copers. These findings offered partial support to our hypotheses, indicating that both individuals with CAI and copers demonstrated distinct characteristics in muscle mechanics and energetics during single leg jump landing in comparison to their healthy controls.

The first important findings of this study were copers exhibited greater peak muscle forces in the vastus lateralis and medial gastrocnemius during landing compared to healthy individuals. The SPM analysis supported this finding, illustrating that copers had higher vastus lateralis and medial gastrocnemius muscle force than healthy controls during 42%–55 % and 21%–43 % of the landing phase, respectively. Similar to the prior research on ankle sprains, suggesting alterations in proximal lower limb biomechanics during sporting activities among individuals with ankle sprains [39]. Our findings reveal that copers exhibited a novel landing pattern- 'knee strategy'. More specifically, being a bi-articular muscle, the primary functions of the medial gastrocnemius include knee flexion and ankle plantarflexion [40]. Research has indicated that the medial gastrocnemius coordinates ankle-knee joint function through muscle synergy, thereby enhancing joint stability and control [41]. Similarly, the vastus lateralis contributes to knee extension and hip flexion, assisting in counteracting lateral forces during landing and restricting excessive leg swinging during lateral movements [42]. Therefore, individuals with copers may achieve a more stable landing by greater activation of the vastus lateralis as well as the medial gastrocnemius. Another study [10] also validated that coper underwent a 're-learning' process of pre-injury movement patterns, restoring a neuromuscular control pattern closer to that of healthy individuals.

The hip, knee, and ankle form an interconnected kinetic chain within the lower extremity, where injuries at the distal ankle could potentially result in abnormal function of the proximal knee and hip [43]. The observed knee strategy among copers might serve as a

Table 3
Detailed results of the mechanical power between those with CAI, copers and healthy controls during single leg jump landing.

Parameters	Group			*F*	*P*-value	*ES*
	CAI Mean (SD)	Copers Mean (SD)	Healthy Mean (SD)			
Total mechanical power (W/kg)	−21.47 ± 1.88	−21.02 ± 2.06	−20.16 ± 2.09	3.563	<0.05	0.511
Ankle mechanical power (W/kg)	−4.42 ± 0.29	−4.52 ± 0.18	−5.82 ± 0.32	268.419	<0.001	0.768
Knee mechanical power (W/kg)	−13.95 ± 1.93	−14.46 ± 2.08	−13.23 ± 2.03	3.009	0.05	0.385
Hip mechanical power (W/kg)	−2.90 ± 0.37	−2.04 ± 0.22	−1.82 ± 0.49	73.782	<0.001	0.633
Ankle contribution to total mechanical power (%)	20.75 ± 2.36	21.72 ± 2.38	25.41 ± 2.81	30.457	<0.001	0.433
Knee contribution to total mechanical power (%)	64.72 ± 4.21	68.49 ± 3.55	65.55 ± 4.12	7.917	0.001	0.080
Hip contribution to total mechanical power (%)	14.53 ± 2.63	9.80 ± 1.51	9.04 ± 2.34	57.942	<0.001	0.685

Note: SD: standard deviation; ES: effect size; CAI: chronic ankle instability.

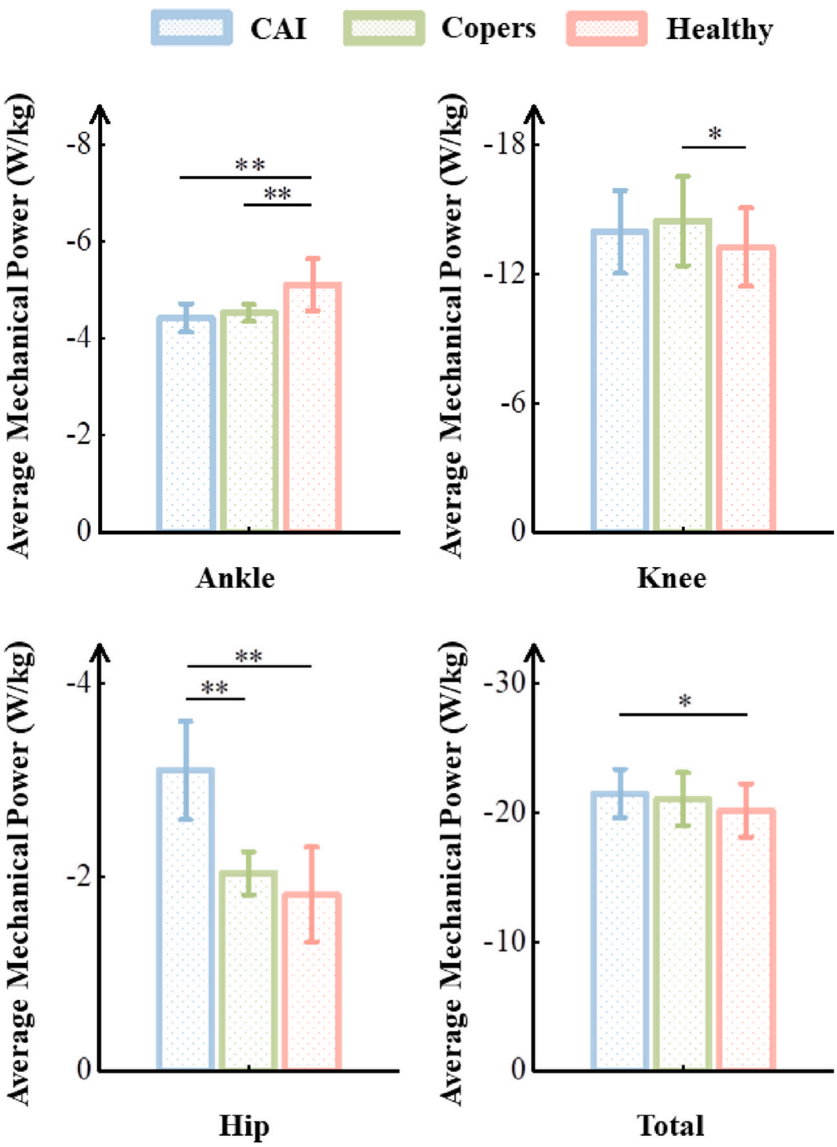

Fig. 5. Mean ± SD average mechanical power in people with CAI, copers and healthy controls during single leg jump landing. *: significant difference with $p < 0.05$. **: significant difference with $p < 0.001$.

compensatory mechanism aimed at compensating ankle deficiencies and maintaining overall body balance. However, this compensation might contribute to the potential risk of anterior cruciate ligament (ACL) injury. This is because most ACL injuries occur during forceful outward contractions of the vastus muscles [44]. Studies have shown that the angle of knee flexion and extension during landing, the point of impact on the ground, and a poorly controlled landing are all closely related to the development of ACL injuries. When individuals change their landing strategy, especially when relying on the knee for too much compensatory activity, it may lead to stress concentrations in the knee joint and increase the ACL stresses [45]. Additionally, abnormal gastrocnemius muscle contractions can lead to a posterior shift in the distal femur, increasing stress on the ACL [46]. Nonetheless, further studies are required for validation. Notably, ACL injury risk during landing might be higher in the CAI individuals, as muscle force of vastus muscles were significantly higher in the CAI during most of the landing phase when compared to the healthy groups. Consistent with prior research [47], our current study observed alterations in proximal hip biomechanics in individuals with CAI during landing. Specifically, we found higher peak muscle forces in the gluteus medius and gluteus maximus among CAI compared to the healthy groups. The biomechanical properties of the hip joint afford it extensive movement capabilities, facilitating more flexible rotation, flexion, and extension in various positions [48]. This extensive range of motion typically suggests enhanced control over body posture in specific situations. Therefore, the identified neuromuscular abnormality in CAI appears to be a feedback mechanism aimed at achieving a more stable landing pattern.

Another notable finding was that a shift of negative mechanical power from distal to the proximal joints during landing of CAI and copers. In addition to ACL injury risk, this compensatory mechanism may lead to long-term effects on joint health, including increased wear and tear on the knee and hip joints, as well as muscle fatigue. Such prolonged compensation may also impair motor control and overall movement efficiency, potentially affecting performance and rehabilitation outcomes in the long run. The transfer of mechanical power from the ankle joint to the knee joint in copers further validates our prior observation regarding the knee strategy employed by copers during landing [49–51]. This finding indicates that ankle sprains may impact the ankle's ability to generate negative mechanical power. Consequently, the increased negative mechanical power at the proximal joints compensates for the decreased generation at the ankle joint during landing. Beck et al. identified that the activation of proximal muscles, such as knee flexors, was perceived to be more energetically demanding when contrasted with the activation of distal muscles, like ankle plantar flexors, owing to the heightened involvement of muscle [52]. The cause of the distal-to-proximal joint work redistribution may be related to higher muscular force generation in the knee extensors and hip flexors, as this musculoskeletal abnormality was found in our current study. Mineta et al. found a correlation between reduced stability in single leg landings after ankle sprains and decreased ankle energy absorption [53]. The mechanisms involving energy redistribution in CAI and copers during landing could potentially play a crucial role in causing secondary sprains. Similar redistribution mechanisms have been observed in studies inducing fatigue [54]. Alexis et al. found [55] an increase in the peak ankle inversion angle during initial landing contact after fatigue, which correlated with an elevated risk of recurrent ankle sprains [56]. Although direct evidence of neural fatigue mechanisms in copers is currently lacking, this potential association holds significant research implications for devising improved rehabilitation programs aimed at preventing secondary sprains. This discovery highlights the potential for ankle rehabilitation robots with exoskeleton assistance designed specifically for copers patients. These systems aim to fulfill the mechanical work requirements during landing, thus delaying the onset of joint work redistribution mechanisms and restoring a normal loading environment.

A further finding of this study was that significant differences in peak muscle force of the soleus and peroneus longus muscles were observed during landing between the copers and healthy individuals. Our findings suggest that copers make slight adjustments in their landing movement patterns, seemingly aimed at mitigating the risk of secondary sprains. Specifically, the soleus muscle, acting as a plantarflexor, works to sustain ankle plantarflexion during landing. It has been suggested that the reduced force production of the soleus muscle in copers may lead to a decrease in the ankle's plantarflexion angle during landing. This reduction in plantarflexion could potentially lower the risk of ankle sprains, as excessive plantarflexion is associated with a higher risk of injury [57]. Wright et al. show that increased ankle plantarflexion angles during landing were associated with a higher risk of ankle sprains. Thus, it's reasonable to hypothesize that the reduced muscle force in copers might result in a decreased ankle plantarflexion angle during landing, consequently lowering the risk of ankle sprains. Regarding the peroneus longus muscle, evidence suggests its role as a primary evertor muscle responsible for shifting the center of pressure (COP) from the lateral to the medial aspect of the body [58]. We observed higher peroneus longus muscle force in CAI and copers individuals. This finding isn't surprising given the peroneus longus muscle's association with abnormal biomechanics related to ankle sprains. Increased peroneus longus muscle force might indicate an adaptive response by the body after initial sprain [59]. This adaptation might involve alterations in movement patterns aimed at enhancing peroneus longus muscle control and mitigating increased ankle inversion due to the substantial ground reaction forces exerted on the lateral aspect of the foot during landing [60].

This study had certain limitations. Firstly, since the participant cohort exclusively comprised males, the outcomes may not be broadly applicable to female individuals with CAI and copers and there may be differences in the findings produced by female subjects versus male subjects. Given this limitation, future research should include other populations, such as female athletes or older individuals to assess potential sex differences in the biomechanical adaptations observed in copers and CAI individuals. Second, in this study, hamstring muscle force wasn't calculated. Considering the significant role played by hamstrings during the landing phase, providing more information could help clinicians and rehabilitation trainers devise a more comprehensive rehabilitation program. A follow-up study will include an exploration of the hamstring muscle. Third, while this study updated the musculotendon parameters in the model using each participant's height, weight, and race, employing individualized musculoskeletal models derived from participant scans could potentially modify the current outcomes and offer deeper insights.

5. Conclusion

The current study investigated muscle mechanics and energetics in individuals with chronic ankle instability (CAI), copers and healthy controls during single leg jump landing. Copers exhibited a novel landing pattern resulting from compensatory mechanisms involving the medial gastrocnemius, vastus lateralis, and soleus muscles. This landing pattern aims for increased stability but potentially elevates the risk of ACL injury. Alterations in muscle mechanics could have influenced the redistribution of joint mechanical power, showing a shift of negative mechanical power from the ankle to the hip in CAI individuals during landing and to the knee in copers. The redistribution of mechanical power at the ankle observed in CAI and copers might correlate with secondary ankle sprains. This study holds significant practical implications for clinicians regarding the development of effective rehabilitation programs.

CRediT authorship contribution statement

Tianle Jie: Writing – original draft, Visualization, Software, Methodology, Investigation, Formal analysis, Conceptualization. **Datao Xu:** Writing – review & editing, Visualization, Supervision, Resources, Methodology, Investigation, Conceptualization. **Huiyu Zhou:** Writing – review & editing, Methodology, Investigation, Conceptualization. **Yongyan Zhang:** Writing – review & editing, Supervision, Resources, Investigation, Conceptualization. **Minjun Liang:** Writing – review & editing, Supervision, Resources, Investigation. **Julien S. Baker:** Writing – review & editing, Supervision, Resources, Methodology, Conceptualization. **Yaodong Gu:** Writing – review & editing, Visualization, Supervision, Software, Methodology, Investigation, Funding acquisition, Conceptualization.

Ethics approval and consent to participate

Ningbo University's Ethics Committee has accepted the study protocol (Approval Number: RAGH20220620), and all subjects supplied and signed written informed permission.

Data availability statement

All data relevant to the current study are included in the article, further inquiries can be directed to the corresponding author.

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Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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