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Original Research

Altered Movement Biomechanics in Chronic Ankle Instability, Coper, and Control Groups: Energy Absorption and Distribution Implications

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Context: Patients with chronic ankle instability (CAI) exhibit deficits in neuromuscular control, resulting in altered movement strategies. However, no researchers have examined neuromuscular adaptations to dynamic movement strategies during multiplanar landing and cutting among patients with CAI, individuals who are ankle-sprain copers, and control participants.

Objective: To investigate lower extremity joint power, stiffness, and ground reaction force (GRF) during a jump-landing and cutting task among CAI, coper, and control groups.

Design: Cross-sectional study.

Setting: Laboratory.

Patients or Other Participants: A total of 22 patients with CAI (age = 22.7 ± 2.0 years, height = 174.6 ± 10.4 cm, mass = 73.4 ± 12.1 kg), 22 ankle-sprain copers (age = 22.1 ± 2.1 years, height = 173.8 ± 8.2 cm, mass = 72.6 ± 12.3 kg), and 22 healthy control participants (age = 22.5 ± 3.3 years, height = 172.4 ± 13.3 cm, mass = 72.6 ± 18.7 kg).

Intervention(s): Participants performed 5 successful trials of a jump-landing and cutting task.

Main Outcome Measure(s): Using motion-capture cameras and a force plate, we collected lower extremity ankle-, knee-, and hip-joint power and stiffness and GRFs during the jump-landing and cutting task. Functional analyses of variance were used to evaluate between-groups differences in these dependent variables throughout the contact phase of the task.

Results: Compared with the coper and control groups, the CAI group displayed (1) up to 7% of body weight more posterior and 52% of body weight more vertical GRF during initial landing followed by decreased GRF during the remaining stance and 22% of body weight less medial GRF across most of stance; (2) 8.8 W/kg less eccentric and 3.2 W/kg less concentric ankle power, 6.4 W/kg more eccentric knee and 4.8 W/kg more eccentric hip power during initial landing, and 5.0 W/kg less eccentric knee and 3.9 W/kg less eccentric hip power; and (3) less ankle- and knee-joint stiffness during the landing phase. Concentric power patterns were similar to eccentric power patterns.

Conclusions: The CAI group demonstrated altered neuromechanics, redistributing energy absorption from the distal (ankle) to the proximal (knee and hip) joints, which coincided with decreased ankle and knee stiffness during landing. Our data suggested that although the coper and control groups showed similar landing and cutting strategies, the CAI group used altered strategies to modulate impact forces during the task.

Key Words: ankle sprains, kinetics, energetics, landing mechanics, ground reaction forces

Key Points

- The chronic ankle instability group demonstrated alterations in lower extremity motor control during initial landing, as evidenced by decreasing ankle-joint energy absorption and increasing knee- and hip-joint energy absorption compared with the ankle-sprain coper and control groups.
- The altered energy patterns coincided with decreased ankle and knee stiffness, which may contribute to a compensatory landing strategy for attenuating increased posterior and vertical ground reaction forces during initial landing.

Lateral ankle sprains (LASs) are the most common musculoskeletal injuries in athletic[1] and nonathletic populations[2] and result in substantial financial[2] and time loss.[1] Furthermore, up to 70% of patients with LASs may develop *chronic ankle instability* (CAI), a condition characterized by perceived instability, recurrent episodes of giving way and sprains, and mechanical or sensorimotor impairments.[3] The sensorimotor impairments associated with CAI include impaired proprioception[4] and neuromuscular control[5] in the lower extremity. These impairments can also have long-term health consequences (eg, osteoarthritis), thereby reducing the quality of life.[6]

In spite of a high recurrence rate of LASs, some individuals do not experience subsequent injuries, residual symptoms, and functional disability; they are known as *ankle-sprain copers*.[7] Comparisons with copers may help us to identify the underlying risk factors for CAI and provide clues for better preventive and rehabilitation strategies for

patients with CAI.[7] In recent studies, researchers have shown sensorimotor deficiencies, involving sensorimotor function[8] and postural control,[9] in patients with CAI relative to copers. Although patients with CAI appeared to have greater alterations in landing mechanics that may predispose them to recurrent LAS,[10] copers displayed relatively small alterations in landing mechanics compared with control participants.[11]

Lateral ankle sprains frequently occur during sports involving jumping, landing, and cutting.[1] However, limited data about 3-dimensional (3D) ground reaction force (GRF), joint power, and joint stiffness are available to help clinicians understand the mechanisms of LASs and lower extremity injuries during landing.[12] Excessive GRF and altered joint placement have been identified as risk factors for lower extremity injury.[13] During landing, GRF is attenuated through the lower extremity musculoskeletal structures, largely by the eccentric action of the plantar flexors and the knee and hip extensors.[14,15] Delahunt et al[10] reported greater and quicker onset of peak vertical and posterior GRFs during initial landing, suggesting that patients with CAI would experience excessive stress on the lower extremity musculoskeletal structures in a shorter period and thereby increase the risk of soft tissue injury.[13] Moreover, sensorimotor impairments are thought to reduce joint stiffness at the ankle during landing, which potentially decreases dynamic joint stability.[16,17] Evaluating joint power may also be important for understanding how patients with CAI modulate impact GRF through the interaction of the lower extremity joints.

Static measures of sensorimotor impairments, including neural activation,[8] plantar cutaneous sensitivity,[18] and postural control,[9] have been compared between CAI and coper groups. However, the generalizability of these data to injuries sustained during movement may be limited. Researchers have examined landing mechanics in patients with CAI compared with copers, controls, or both,[19,20] but the landing tasks were simple, uniplanar movements with no directional changes (ie, cutting) and may not represent the mechanisms of LASs, given that most sprains occurred during landing (45%) and cutting (30%).[21] We believed that using a dynamic, physically demanding task involving a single-legged, high-impact deceleration landing followed by a rapid side-cutting acceleration jump would better represent the movements often associated with ankle-sprain mechanisms. Therefore, the purpose of our study was to examine biomechanical factors (eg, 3D GRF, joint power, and joint stiffness) among CAI, coper, and control groups during a maximal single-legged jump-landing and cutting task. We hypothesized that the CAI group would exhibit greater GRF and less joint power and stiffness during the task than the coper and control groups.

METHODS

Participants

A total of 66 physically active college students participated in this study. We defined *physically active* as exercising at least 3 days per week for a total of 90 minutes during the 3 months before the study. Sample size was calculated according to previous data[19] using a Cohen d effect size of 0.69 and β of .2. All participants were assigned to the CAI, ankle-sprain coper, or control group on the basis of previous studies.[3,7] The involved ankles of the coper and control groups were matched to the involved ankles of the CAI group. Inclusion criteria for the CAI group were based on a position statement of the International Ankle Consortium (IAC)[3] and were described in a previous study.[11] The inclusion criteria for the coper and control groups have also been described.[11] The exclusion criteria for all groups were based on a position statement of the IAC[3] and were presented in a previous article.[11] Participant demographics are shown in Table 1. All participants provided written informed consent, and the study was approved by the Brigham Young University Institutional Review Board for Human Subjects.

Experimental Procedure

All participants dressed in spandex clothing (model HeatGear; UnderArmour, Baltimore, MD) and athletic shoes (model T-Lite XI; Nike, Beaverton, OR). Anthropometric characteristics (ie, age, height, mass, and shank length) were measured for each participant. Recruits performed a 5-minute warm-up by treadmill walking at a self-selected pace. Fifty-nine reflective markers were placed over anatomic landmarks,[22] including the head, C7, acromion processes, inferior angles of the scapulae, T7, lateral epicondyles of the humeri, heads of the ulnae, sternum, anterior- and posterior-superior iliac spines, greater trochanters, medial and lateral femoral condyles, medial and lateral malleoli, posterior heels, dorsal midfeet, and medial and lateral feet and between the second and third metatarsal heads of both feet. Four rigid clusters with 4 markers were placed bilaterally on the lateral thigh and shank. Twelve high-speed video cameras (model T10S; Vicon Motion Systems Ltd, Oxford, UK) recording at 240 Hz were used to collect 2 dynamic video trials so that we could calculate right and left functional hip-joint centers.[11] Participants completed the jump-landing and cutting task as described by Son et al.[11]

Data Analysis

Dependent variables were analyzed and normalized to 100% of the *stance phase*, which was defined as the time from initial contact (1%) to toe-off (100%) with a 25-N vertical GRF threshold. Stance was divided into 2 phases: (1) landing (eccentric), which was from initial contact (IC) to peak dorsiflexion (at an average 50% of stance), knee flexion (at an average 50% of stance), and hip flexion (at an average 32% of stance), and (2) cutting (concentric), which was from peak sagittal-plane ankle, knee, and hip angles to toe-off. The concentric cutting phase of the hip joint was initiated at 32% of stance, which was attributed to the characteristics of our task in that the hip initiated a directional change earlier than the ankle and knee.

Three-dimensional trajectories for each reflective marker and the GRF data were identified using Vicon Nexus (version 2.4.0; Vicon Motion Systems Ltd) and exported to Visual 3D (version 5; C-Motion, Inc, Germantown, MD). The trajectory and GRF data were filtered using a fourth-order low-pass Butterworth filter with a cutoff frequency of 10 Hz. As described in a previous study,[11] a rigid link model (foot, shank, thigh, and pelvis segments) was created, and ankle-, knee-, and hip-joint angles were calculated using a Cardan rotation sequence (X, Y′, Z′′).

Table 1. Participant Demographics

Characteristic	Group		
	Chronic Ankle Instability	Coper	Control
Sex (male/female), n	14/8	14/8	14/8
	Mean ± SD		
Age, y	22.7 ± 2.0	22.1 ± 2.1	22.5 ± 3.3
Height, cm	174.6 ± 10.4	173.8 ± 8.2	172.4 ± 13.3
Mass, kg	73.4 ± 12.1	72.6 ± 12.3	72.6 ± 18.7
Foot and Ankle Ability Measure, %			
Activities of Daily Living subscale	81.9 ± 7.3	100.0 ± 0.0	100.0 ± 0.0
Sports subscale	60.9 ± 11.6	100.0 ± 0.0	100.0 ± 0.0
Modified Ankle Instability Index, No. of *yes* responses[a]	3.4 ± 1.1	0.0 ± 0.0	0.0 ± 0.0
Ankle sprains, No.	4.1 ± 2.8	2.0 ± 1.1	0.0 ± 0.0

[a] Questions 4–8.

The ankle-joint angles were determined as the rotation of Cardan angles of the foot relative to the tibia in the order of plantar flexion-dorsiflexion (x-axis), internal-external rotation (y-axis), and inversion-eversion (z-axis). At the knee, the order of Cardan angles rotation was flexion-extension (x-axis), valgus-varus (y-axis), and internal-external rotation (z-axis). At the hip, the order of Cardan angles rotation was flexion-extension (x-axis), adduction-abduction (y-axis), and internal-external rotation (z-axis).[11] Joint power was calculated as the product of the net internal joint moment and joint angular velocity.[14] Eccentric power was considered to be kinetic energy absorption, whereas concentric power was considered to be kinetic energy generation. Sagittal-plane ankle-, knee-, and hip-joint stiffness was calculated as the change in net internal moment divided by the angular displacement between IC and peak dorsiflexion as well as knee and hip flexion during landing.[23] Joint power and stiffness were normalized to each participant's body mass (in kilograms). The GRF was normalized to body weight (BW), which is equivalent to the product of body mass and gravity. All GRFs were expressed as a percentage of BW.

Statistical Analysis

Functional analyses of variance (FANOVA; version 2.15.1; R program, RStudio, Boston, MA) were used to comprehensively evaluate between-groups differences and 95% confidence intervals (CIs) for 3D GRF and joint power across the entire stance phase of the jump-landing and cutting task as described in previous studies.[11,24] We plotted our estimates of pairwise comparison functions between groups and 95% CIs to determine differences. Between-groups comparisons were considered different if an estimate of the 95% CI did not include zero. A 1-way analysis of variance was performed to assess between-groups differences for the maximal vertical-jump height and ankle-, knee-, and hip-joint stiffness during landing using JMP Pro 13 software (SAS Institute Inc, Cary, NC). When we found a group difference, we conducted a post hoc Tukey-Kramer honestly significant difference test. The α level was set at .05.

RESULTS

We observed no between-groups differences in maximal vertical-jump height during the jump-landing and cutting task ($F_{2,327} = 0.8089$, $P = .45$). The mean maximal vertical-jump heights were 42.21 ± 13.4 cm, 43.99 ± 11.0 cm, and 42.27 ± 10.6 cm for the CAI, coper, and control groups, respectively.

Ground Reaction Force

The between-groups differences for GRF across the stance phase of the jump-landing and cutting task are shown in Figure 1. Relative to the coper group, the CAI group had up to (1) 22% of BW less medial GRF during 12% to 80% of eccentric and concentric stances; (2) 7% of BW more and 12% of BW less posterior GRF during 12% to 22% of eccentric stance and 26% to 76% of eccentric and concentric stances, respectively; and (3) 52% and 9% of BW more vertical GRF during 4% to 14% of eccentric stance and 82% to 96% of concentric stance, respectively, and 57% of BW less vertical GRF during 18% to 75% of eccentric and concentric stances ($P < .05$). Relative to the control group, the CAI group had up to (1) 8% and 2% of BW more medial GRF during 0% to 10% of eccentric stance and 88% to 95% of concentric stance, respectively, and 16% of BW less medial GRF during 15% to 78% of eccentric and concentric stances; (2) 4% of BW more posterior GRF during 0% to 7% of eccentric stance and 19% of BW less posterior GRF during 24% to 100% of eccentric and concentric stances; and (3) 75% and 6% of BW more vertical GRF during 0% to 15% of eccentric stance and 86% to 100% of concentric stance, respectively, and 53% of BW less vertical GRF during 17% to 77% of eccentric and concentric stances ($P < .05$). Compared with the control group, the coper group had up to (1) 5% and 7% of BW more medial GRF during 0% to 10% and 22% to 36% of eccentric stance, respectively; (2) 5% of BW more and 9% of BW less posterior GRF during 0% to 8% of eccentric stance and 38% to 100% of eccentric and concentric stances, respectively; and (3) 26% of BW more vertical GRF during 0% to 10% of eccentric stance ($P < .05$).

Joint Power

The between-groups differences for ankle-, knee-, and hip-joint power across stance of the jump-landing and cutting task are provided in Figure 2. Relative to the coper group, the CAI group produced up to (1) 8.8 W/kg less eccentric ankle power during 9% to 49% and 3.2 W/kg less

Table 2. Sagittal-Plane Ankle-, Knee-, and Hip-Joint Stiffness During Landing

Stiffness Location	Group, Nm/kg/° (Mean ± SD [95% Confidence Interval])			$F_{2,327}$ Value	P Value
	Chronic Ankle Instability	Coper	Control		
Ankle	0.049 ± 0.013 (0.046, 0.052)	0.054 ± 0.014 (0.051, 0.057)	0.054 ± 0.014 (0.051, 0.056)	4.1298	.02[a,b]
Knee	0.043 ± 0.019 (0.039, 0.047)	0.061 ± 0.019 (0.057, 0.065)	0.059 ± 0.018 (0.056, 0.062)	30.3988	.001[a,b]
Hip	0.157 ± 0.144 (0.129, 0.185)	0.149 ± 0.166 (0.117, 0.182)	0.138 ± 0.144 (0.111, 0.166)	0.4001	.67

[a] Indicates difference between chronic ankle instability and coper groups ($P < .05$).
[b] Indicates difference between chronic ankle instability and control groups ($P < .05$).

concentric ankle power during 53% to 88% of stance ($P < .05$); (2) 6.4 W/kg more and 5.0 W/kg less eccentric knee power during 8% to 18% and 21% to 41% of stance, respectively, and 2.3 W/kg less and 1.7 W/kg more concentric knee power during 50% to 72% and 80% to 93% of stance, respectively ($P < .05$); and (3) 4.8 W/kg more and 3.9 W/kg less eccentric hip power during 8% to 18% and 20% to 28% of stance, respectively, and 2.4 W/kg less and 1.4 W/kg more concentric hip power during 32% to 45% and 71% to 91% of stance, respectively ($P < .05$). Compared with the control group, the CAI group produced up to (1) 3.5 W/kg more and 7.9 W/kg less eccentric ankle power during 0% to 8% and 10% to 50% of stance, respectively, and 4.1 W/kg less concentric ankle power between 56% and 89% of stance ($P < .05$); (2) 8.1 W/kg more and 8.5 W/kg less eccentric knee power during 0% to 18% and 20% to 38% of stance, respectively, and 2.9 W/kg less concentric knee power during 54% to 77% of stance ($P < .05$); and (3) 5.5 W/kg more and 7.0 W/kg less eccentric hip power during 8% to 17% and 20% to 30% of stance, respectively, and 3.5 W/kg less and 2.0 W/kg more concentric hip power during 32% to 52% and 65% to 92% of stance, respectively ($P < .05$). Relative to the control group, the coper group produced up to (1) 2.5 W/kg more and 0.8 W/kg less eccentric ankle power during 0% to 10% and 30% to 50% of stance, respectively, and 1.2 W/kg less concentric ankle power during 80% to 90% of stance ($P < .05$); (2) 2.1 W/kg more and 3.5 W/kg less eccentric knee power during 0% to 12% and 21% to 31% of stance, respectively, and 0.8 W/kg more and 1.5 W/kg less concentric knee power during 50% to 55% and 67% to 81% of stance, respectively ($P < .05$); and (3) 3.0 W/kg less eccentric hip power during 19% to 30% of stance and 1.9 W/kg less concentric hip power during 36% to 53% of stance ($P < .05$).

Joint Stiffness

Ankle- and knee-joint stiffness during landing differed among groups ($P < .05$; Table 2). Tukey-Kramer HSD post hoc tests showed that the CAI group had less ankle- and knee-joint stiffness than both the coper and control groups. We observed no between-groups differences in hip-joint stiffness.

DISCUSSION

The purpose of our study was to determine 3D GRF and joint-power patterns and joint stiffness during a novel, dynamic jump-landing and cutting task among CAI, coper, and control groups. Our main findings were that, although the coper and control groups demonstrated similar or relatively small differences in landing and cutting mechanics, the CAI group exhibited (1) more posterior and vertical GRFs during initial landing and fewer medial, posterior, and vertical GRFs during midlanding and cutting; (2) more eccentric knee and hip power during initial landing and less eccentric ankle, knee, and hip power during midlanding, which resulted in less concentric ankle, knee, and hip power during early cutting and more concentric knee and hip power during terminal cutting; and (3) less ankle- and knee-joint stiffness than the coper and control groups.

Relative to the coper and control groups, the CAI group exhibited greater posterior and vertical GRF during initial landing followed by decreased GRF during the remaining stance. These results were consistent with those of a previous study[13] in which patients with CAI displayed more vertical and posterior GRFs during initial landing than controls. Increases in initial impact force could be attributed to movement alterations after ankle injury due to altered feedforward motor control.[25] Rowley and Richards[26] reported that individuals with greater dorsiflexion angles appeared to have higher vertical GRF at IC during a drop landing. Our findings of higher vertical and posterior GRFs could be due to altered ankle kinematics, as noted in a previous study[11] of the same CAI cohort in which patients with CAI demonstrated greater dorsiflexion (5.6°–7.4°) from IC to initial landing than did copers and controls.[11] Son et al[11] suggested that patients with CAI may voluntarily try to maintain a safe landing strategy using a dorsiflexed (closed-packed) position during initial landing to compensate for self-perceived instability. Our data indicated that, after the initial landing phase, the CAI group displayed decreased medial, posterior, and vertical GRFs compared with the coper and control groups. Researchers[14,27] have shown a relationship between knee-flexion angle and vertical GRF during landing. Active hip- and knee-flexion kinematics appeared to reduce impact GRFs during landing.[14] Our results of reduced vertical and posterior GRFs could reflect greater angular displacement at the knee (6.3° more flexion) and hip (5.6° more flexion), as supported by previous data,[11] which may have led to reduced knee-joint stiffness in patients with CAI. Our data are supported by those of previous studies[14,27] in which the authors found that decreased knee-joint stiffness was associated with reduced vertical GRF during a landing task. From an injury-prevention standpoint, greater knee

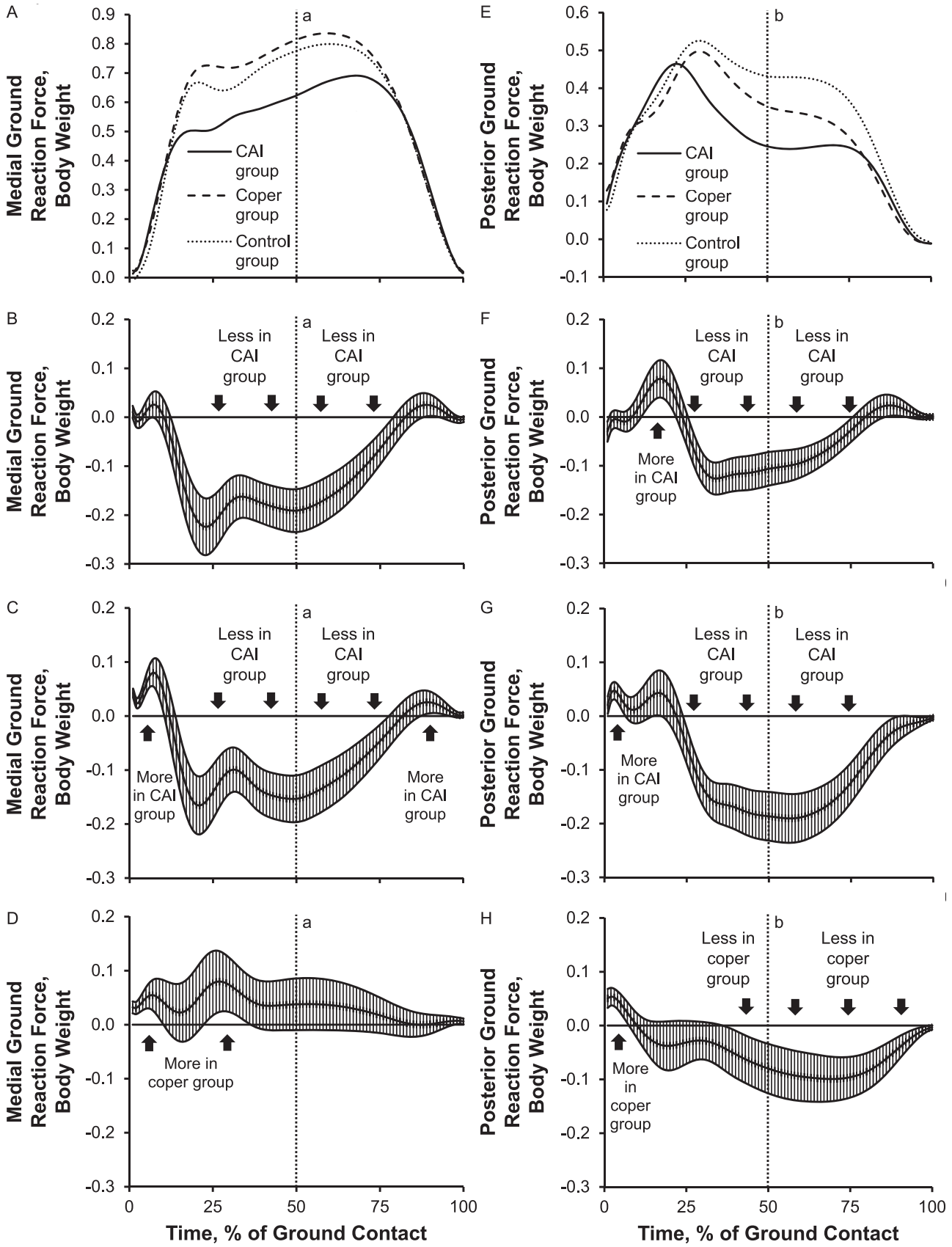

Figure 1. Three-dimensional ground reaction force patterns across the entire stance phase during the jump-landing and cutting task. Medial ground reaction force: A, of the chronic ankle instability (CAI), coper, and control groups during stance phase; B, between the CAI and coper groups; C, between the CAI and control groups; D, between the coper and control groups. Posterior ground reaction force: E, of the CAI, coper, and control groups during stance phase; F, between the CAI and coper groups; G, between the CAI and control groups; H, between the coper and control groups. Continued on next page.

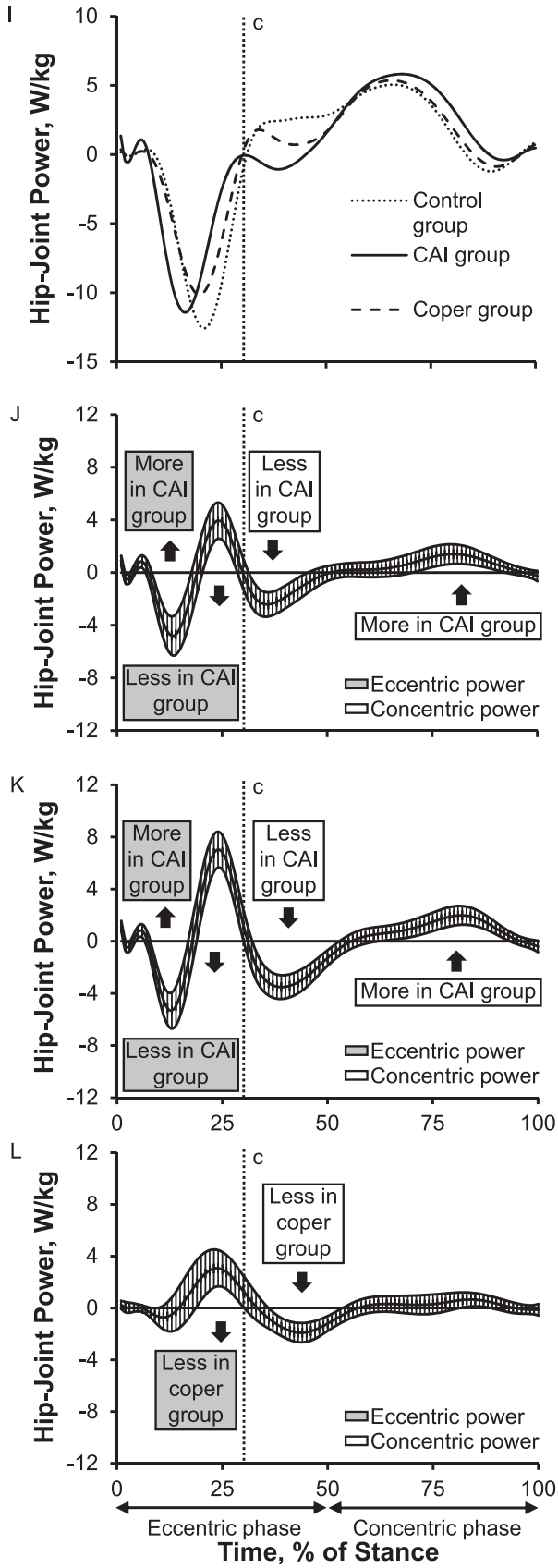

Figure 1. Continued from previous page. Vertical ground reaction force: I, of the CAI, coper, and control groups during stance phase; J, between the CAI and coper groups; K, between the CAI and control groups; L, between the coper and control groups.

and hip flexion during landing could help dissipate impact GRF over a longer period of time, thereby reducing the risk of lower extremity injury.[15] Our patients with CAI may have attempted to reduce impact forces after the initial landing phase by using a soft-landing strategy to prevent soft tissue injury. However, given that the lower extremity is connected by a kinetic-kinematic chain, all 3 joints should be considered together because researchers[11] have suggested that greater knee- and hip-flexion angles may result from limited dorsiflexion (2.5°) during the jump-landing and cutting task.

Compared with the coper and control groups, the CAI group displayed less medial GRF across stance. Changes in medial GRF during landing provide important information for identifying risk factors for LAS related to joint placement and the center-of-mass (COM) position. In the same cohorts, Son et al[11] reported that patients with CAI landed in up to 6° less hip abduction in the frontal plane than copers.[11] They concluded that patients with CAI positioned their COM closer to the center of pressure (COP) on the landing foot by decreasing the hip-abduction angle and increasing the hip internal-abduction moment during a jump-landing and cutting task compared with copers. The jump-landing and cutting task was highly demanding and challenging, requiring a sudden deceleration of the involved limb followed by a rapid acceleration to the side of the contralateral limb. Considering the nature of this dynamic task, it may be easier and safer for patients with CAI to stabilize the downward motion of their COM closer to the COP in the frontal plane, leading to a less abducted hip position. This upright position of the thigh during landing and cutting may be a self-protective landing strategy, resulting in less medial GRF during stance, as we observed. Although a more vertically upright position of the thigh could increase the demands of hip-joint stability during landing, it may move the COP closer to the lateral border of the stance foot (ie, laterally deviated COP), causing a greater external inversion moment during the cutting phase. During the lateral-cutting movement, the external inversion moment could occur by creating a moment arm between the subtalar joint axis and the GRF.[28] Therefore, the frontal plane should be considered by clinicians who try to reduce the injury risk through prevention and rehabilitation strategies.

Relative to the coper group, the CAI group demonstrated less ankle-joint power throughout most of stance. Given that the plantar flexors contribute to absorbing impact GRF during landing,[14] eccentric plantar-flexion strength is important. Deficits in plantar-flexion strength have been reported in patients with CAI.[29] Although we did not measure strength, our findings of reduced eccentric and concentric ankle-joint power were consistent with reported strength deficits in patients with CAI.[16,29] In fact, ankle-joint stiffness has been reported to be lower during running in patients with CAI than in controls.[17] Altered ankle stiffness could be attributed to sensorimotor deficits in the

←
B–D, F–H, and J–L, mean differences (bold solid curve) and corresponding 95% confidence intervals (shaded area). When the shaded area does not overlap with the zero line (horizontal line), a difference is indicated between groups ($P < .05$). [a] Peak dorsiflexion angle (50% of stance). [b] Knee-flexion angle (50% of stance). [c] Hip-flexion angle (32% of stance).

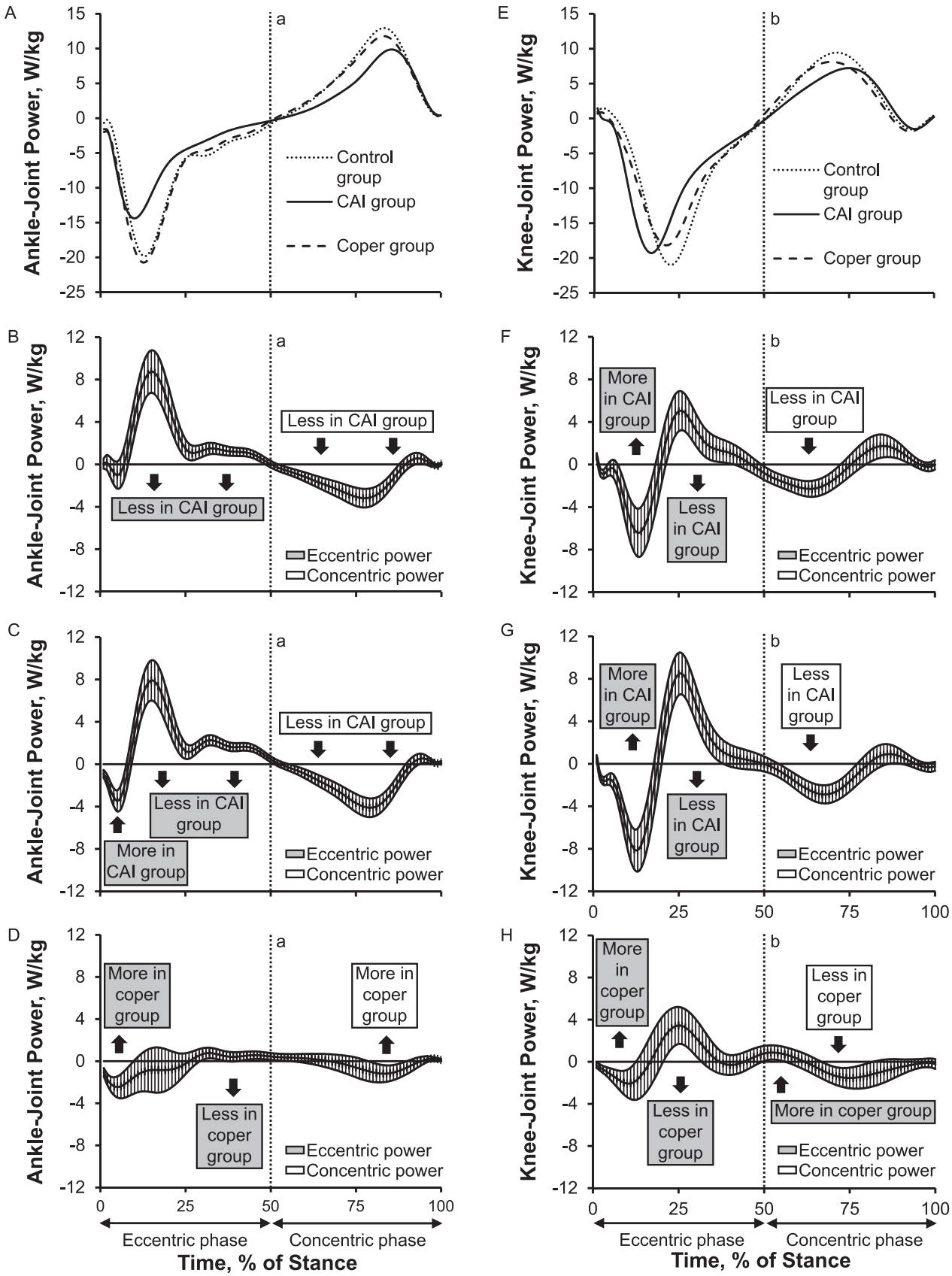

Figure 2. Lower extremity energy patterns across the entire stance phase during the jump-landing and cutting task. Ankle-joint power: A, of the chronic ankle instability (CAI), coper, and control groups during the stance phase; B, between the CAI and coper groups; C, between the CAI and control groups; D, between the coper and control groups. Knee-joint power: E, of the CAI, coper, and control groups during stance phase; F, between the CAI and coper groups; G, between the CAI and control groups; H, between the coper and control groups. **Continued on next page.**

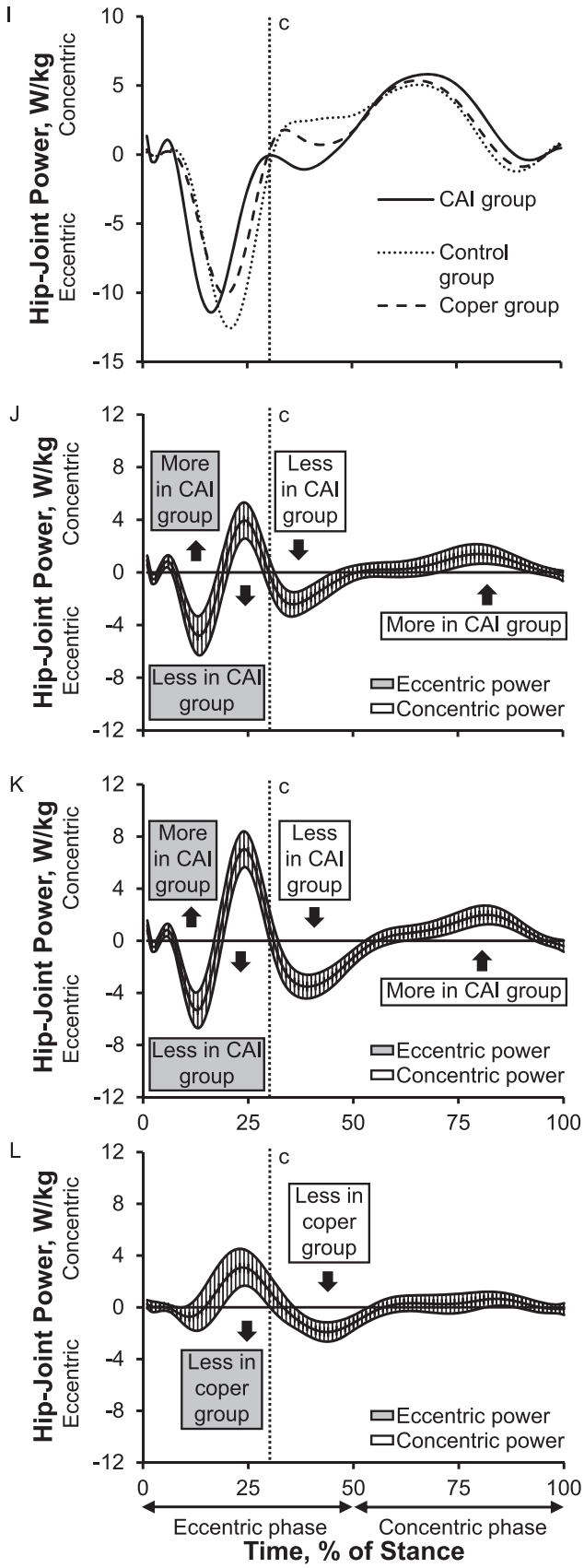

Figure 2. Continued from previous page. Hip-joint power: I, of the CAI, coper, and control groups during stance phase; J, between the CAI and coper groups; K, between the CAI and control groups; and, L, between the coper and control groups. B–D, F–H, and J–L, mean differences (bold solid curve) and corresponding 95% confidence intervals (shaded area). When the shaded area does not overlap with the zero line (horizontal line), a difference is indicated between groups ($P < .05$). [a] Peak dorsiflexion angle (50% of stance). [b] Knee-flexion angle (50% of stance). [c] Hip-flexion angle (32% of stance).

ankle.[4] Because joint stiffness is required to resist an applied external force, such as GRF,[27] less joint stiffness could diminish dynamic joint stability. Our data suggested that, relative to the coper group, reductions in plantar-flexor energy absorption and ankle-joint stiffness with greater vertical and posterior GRF in the CAI group during the initial landing may have increased the ground impact forces being transmitted to the soft tissues around the ankle joint.[30]

The CAI group appeared to increase knee- and hip-joint power during initial landing and then decrease joint power during the remaining landing phase. The knee extensors are important in attenuating impact force via eccentric action during landing.[27] Compared with the coper group, the CAI group increased its reliance on the hip joint during the single-legged landing[20] due to its mechanical advantages (eg, greater muscle volume and strength).[27] Hass et al[31] suggested that patients with CAI altered their spinal and supraspinal motor-control mechanisms. Our findings of altered patterns of knee- and hip-joint power, ie, absorbing more impact GRF using greater eccentric knee- and hip-joint power during the initial landing, may indicate a self-protective mechanism to compensate for reduced ankle-joint power. Patients with CAI may develop new motor programming (an intralimb-reweighting landing strategy), redistributing the load from the weakened distal ankle to the proximal knee and hip joint to effectively attenuate increased GRF and control deceleration of the COM. More data are needed to determine whether these compensation strategies are the key to perpetuating the chronic nature of ankle instability.

The large differences in joint power and GRF we noted were due not only to the amplitudes but also to the timing of the peaks. When observing the greatest differences (Figures 1 and 2) among the groups, a change in how much joint power or GRF exists among groups was evident, but the CAI group presented initial peaks earlier in the time spectrum, accounting for some of the differences in amplitude because the differences existed across time. Considering this observation when interpreting the large-amplitude differences presented in the results could be important. Furthermore, the earlier onset and faster absorption of loads in the CAI group have potential clinical implications.

Limitations

Our study had several limitations. Given that the study design was cross-sectional, it remains unclear whether the observed alterations in landing and cutting movement strategies among the CAI group were due to an adaptive motor-control alteration to an ankle-sprain injury or were present before the injury. Future prospective studies are needed to clarify a causal effect of CAI on movement strategies during landing. We used self-reported questionnaires, including the Foot and Ankle Ability Measure and Modified Ankle Instability Index, to differentiate patients with CAI from copers and controls. Although the authors of

most recent studies have followed the inclusion criteria for patients with CAI endorsed by the IAC,[3] they have used slightly different self-report questionnaires to identify these patients, which may have resulted in the varied findings among studies. In addition, mechanical ankle instability may have affected movement patterns.[19] However, mechanical ankle instability was not an inclusion criterion in our study, which may have led to conflicting results regarding between-groups differences in movement patterns.

Practical Implications

To our knowledge, we are the first to comprehensively analyze the energy patterns, joint stiffness, and GRFs of the entire lower extremity during a multiplanar jump-landing and cutting task among CAI, coper, and control groups. The CAI group displayed less ankle-joint power and greater vertical and posterior GRFs during initial landing than the coper and control groups. The CAI group demonstrated earlier onsets of peak eccentric ankle power and peak vertical and posterior GRFs than the coper and control groups. This decreased time for shock absorption (shorter time of peak eccentric ankle power) during a greater loading rate (shorter time to peak GRF) likely increased stress at the ankle joint. These altered GRF and energy patterns may increase mechanical loading at the ankle, which could predispose this patient population to loads related to recurrent LASs. In the long term, increased joint loading may also accelerate the development of osteoarthritis in patients with CAI.[6]

Individuals with CAI demonstrated a different motor-control strategy by redistributing the impact force from the distal to the proximal joints during initial landing. However, in a recent prospective study, De Ridder et al[32] reported that decreased hip-extension strength was a risk factor for LAS; patients with CAI have shown deficits in knee-extensor and -flexor torques.[29] Our CAI group exhibited less knee-joint stiffness during landing. If patients with CAI adopt a landing strategy that is proximal-joint dominant without regaining knee and hip strength, they may be more susceptible to lower extremity injury, including ankle sprains, in the long term.[31] For these reasons, rehabilitation exercises for patients with CAI should focus not only on the ankle but also on the proximal joints (knee and hip) to increase joint stiffness and power. Furthermore, interventions should emphasize reprogramming the patient's landing strategy to more evenly distribute impact forces throughout the entire lower extremity.

CONCLUSIONS

The CAI group demonstrated alterations in lower extremity motor control during initial landing, as evidenced by decreasing energy absorption at the ankle and increasing knee- and hip-joint energy absorption compared with the coper and control groups. These altered energy patterns coincided with decreased ankle and knee stiffness, which may contribute to a compensatory landing strategy for attenuating increased posterior and vertical GRFs during initial landing.

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