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# Peak Forces and Force Generating Capacities of Lower Extremity Muscles During Dynamic Tasks in People with and Without Chronic Ankle Instability

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# ABSTRACT

People with chronic ankle instability (CAI) exhibit neuromuscular deficits. However, no study has investigated deficits in forces or force-generating capacities of individual muscles in people with CAI during dynamic tasks. Therefore, the purpose of this study was to estimate and compare peak forces and force-generating capacities of individual muscles during dynamic tasks in people with CAI and healthy controls (CON). Eleven people with CAI and eleven CON performed landing, anticipated cutting, and unanticipated cutting as motion capture, force plate, and electromyography data were recorded. A musculoskeletal model was used to estimate the force and force-generating capacity of lower extremity muscles. People with CAI exhibited greater gluteus maximus force and force-generating capacity than CON during all tasks. In addition, people with CAI exhibited greater force-generating capacity of the vastii muscles than CON during the unanticipated cutting task. These findings suggest that, during dynamic tasks, people with CAI exhibit a neuromuscular control strategy that is characterised by differences in peak forces and force-generating capacities of proximal muscles, which may allow them to compensate for previously described deficits in distal muscles.



# Keywords

Biomechanics; musculoskeletal; injury; cutting; OpenSim


# Introduction

Ankle ligament sprains are common musculoskeletal injuries, and people who sprain their ankle ligaments may develop chronic ankle instability (CAI) (Gribble et al.,). Based on the current model of CAI and its causes (Hertel & Corbett,), the deficits associated this pathology can be categorised into three main categories of impairments i.e., pathomechanical (e.g., ankle joint laxity (Hubbard-Turner, ), limited dorsiflexion (Yoon et al., )), sensory-perceptual (e.g., impaired proprioception (Willems et al.,)), and motor-behavioural impairments (e.g., delayed muscle activation (Flevas et al.,), diminished H-reflex (H. Kim et al.,)). Furthermore, recent studies reported people with CAI have a substantially higher risk of developing ankle osteoarthritis because the aforementioned deficits can damage the articular surfaces in the talocrural joint (Valderrabano et al.,). Although researchers and clinicians have developed rehabilitation protocols to combat the negative effects of CAI, rehabilitation outcomes are not always successful (O'Driscoll & Delahunt; Song et al., Tsikopoulos et al., Vallandingham et al.,), which suggests that the mechanisms and deficits in neuromuscular function are not fully understood or that they are not being adequately targeted within rehabilitation programmes.

Previous studies have investigated neuromuscular characteristics in people with CAI with a variety of experimental research methods (Feger et al., K. Kim et al., Simpson et al., Suttmiller & McCann, Willems et al., Wisthoff et al.,). For example, electromyography (EMG) has been used to record and compare muscle activation between people with CAI and matched controls (Feger et al., K. Kim et al., Simpson et al., Suttmiller & McCann,). Results from these studies suggest that people with CAI exhibited greater muscle activation of tibialis anterior during a side-cutting task (Simpson et al.,) and greater muscle activation of medial gastrocnemius during jump landing-cutting motions (K. Kim et al.,). Furthermore, EMG recordings of electrically evoked potentials revealed that people with CAI exhibit a greater decrease in spinal reflex excitability (i.e., Hoffmann reflex) of the soleus when transitioning from bipedal to unipedal stance (Suttmiller & McCann,). In addition, investigations of muscle activation timing relative to initial contact during walking revealed that people with CAI exhibited earlier activations in tibialis anterior, peroneus longus, lateral gastrocnemius, rectus femoris, biceps femoris, and gluteus medius muscles than people without CAI (Feger et al.,). Moreover, dynamometry has been used to investigate strength of ankle muscles in people with CAI (Willems et al., Wisthoff et al.,). For example, people with CAI exhibited lower concentric strength of plantar flexor (Wisthoff et al.,) and lower muscle strength of evertor during isokinetic contraction test (Willems et al.,). Although neuromuscular differences between people with CAI and healthy controls are well characterised, less is known about the function of individual muscles as people with CAI perform dynamic tasks.

Although studies have investigated muscle activation and function via EMG in people with CAI, no previous studies have investigated the force-length-velocity behaviour of individual muscles during functional dynamic tasks, such as jumping or cutting, in this population. A major obstacle for experimental studies that use EMG is that they do not provide information about the forces and length changes of multiple muscles during dynamic tasks. Computer simulations and musculoskeletal modelling, however, provide tools to estimate the kinematics and kinetics of individual muscles. These tools allow for the dynamic estimation of a muscle's peak force, and even its force-length $(f_L)$ and force-velocity $(f_v)$ behaviour during any given task as long as the instantaneous joint kinematics (e.g., joint angle and angular velocity) are also known (Arnold et al.,). Further, this information can also be used to estimate a muscle's instantaneous force-generating capacity, which allows us to better understand the 'relative' ability or capacity of a muscle to produce force and thus complements the analysis of 'absolute' peak muscle forces, which is more traditional.

Therefore, the purpose of this study was to estimate the forces and force-generating capacity of individual lower extremity muscles and to compare these estimates between people with and without CAI during landing and cutting tasks. We hypothesised that the peak muscle forces and force-generating capacity would differ between groups and that these differences would be task-dependent.

# Methods

Twenty-two participants (11 healthy people and 11 CAI) participated in this study (Table 1). A modified ankle instability instrument was used for inclusion and exclusion criteria for the CAI group by quantifying history of ankle sprains and symptoms of the ankle joint (Kipp & Palmieri-Smith, McVey et al.,). Nine questions in the questionnaire asked each participant about the history of ankle sprain and symptoms of CAI, which were inclusion criteria, whereas two questions in the questionnaire asked history of other injuries where participants who reported any fracture and/or bilateral ankle sprain were excluded (Kipp & Palmieri-Smith, McVey et al.,). In addition, the Foot & Ankle Disability Index (FADI) and FADI-Sports questionnaire were used to quantify general and sports-related functional deficits, respectively (Hale & Hertel,). People in the control group (CON) were matched by sex, age, height, weight, and physical activity level, which was quantified via Tegner scores. This study was designed as a cross-sectional study.

Table 1. Demographic information. (CAI: chronic ankle instability group, CON: control group, FADI: Foot & Ankle Disability Index, FADIS: Foot & Ankle Disability Index in Sports, Tegner: Tegner's score)

| Group | Year | Height (m) | Weight (kg) | FADI (%) | FADIS (%) | Tegner |
|---|---|---|---|---|---|---|
| CAI | 22.1 ± 3.2 | 1.68 ± 0.11 | 69.0 ± 19.1 | 90.3 ± 9.4 | 88.6 ± 9.1 | 5.3 ± 1.2 |
| CON | 22.6 ± 4.2 | 1.74 ± 0.11 | 66.8 ± 15.5 | 100 ± 0.0 | 100 ± 0.0 | 5.3 ± 1.0 |

The participants performed three tasks (landing, anticipated cutting, unanticipated cutting) with 32 reflective skin markers attached to their pelvis (anterior superior iliac spine, posterior superior iliac spine, iliac crest), femur (greater trochanter, medial and lateral epicondyle, anterior thigh), tibia (fibular head, lateral shank, medial and lateral malleoli), and foot (calcaneal tuberosity, 1st metatarsal base and head, 5th metatarsal head) (Brown et al., Kipp & Palmieri-Smith, ) and 5 EMG electrodes attached over the soleus, fibularis longus, tibialis anterior, medial gastrocnemius, and lateral gastrocnemius muscles. For the landing task (LAND), participants were asked to perform a forward-jump over a 15 cm box and land on a force plate on a single leg. The forward-jump distance was set to the participant's leg length. For the anticipated and unanticipated cutting tasks, participants performed the same forward jump, but subsequently also performed a 90° cut immediately after landing on the force plate. The cutting direction was indicated with a light stimulus, which was turned on 5 sec before jumping during the anticipated condition (ANT). For unanticipated cutting (UNANT) the light stimulus came on when participants passed through a light beam that was set halfway between the starting position and the force plate. Each participant performed the three tasks in the same order: 1) landing, 2) anticipated cutting, and 3) unanticipated cutting. Thus, the difficulty of tasks was continuously increased. Three-dimensional position of markers, muscle activations, and ground reaction forces (GRF) were collected with motion capture cameras (ViconMx, CA, USA), an EMG system (Bagnoli, Delsys, MA, USA), and a force platform (Advanced Medical Technologies Inc., MA, USA). Sampling frequencies were set to 240 Hz for the cameras and to 1200 Hz for the EMG system and force platform.

Three-dimensional marker positions and GRFs were filtered with lowpass Butterworth filters at cut-off frequencies of 12 Hz. The EMG data were bandpass-filtered with Butterworth filters at cut-off frequencies of 20 and 450 Hz. The filtered EMG data were rectified and smoothed with a lowpass Butterworth filter at cut-off frequency of 10 Hz to obtain an EMG envelope. Then, each EMG signal envelope was normalised by the maximum value of the signal. All data were time-normalised to 0–100 % of task duration (i.e., 200 ms for the landing task and stance phase for cutting tasks). We determined the initial contact of all tasks as the moment when vertical GRF was greater than 10 N. In addition, we determined the end of the stance phase for landing tasks as 200 ms after the initial contact, while we determined the end of the stance phase for cutting tasks as the last time point when vertical GRF was greater than 10 N.

A musculoskeletal model with 23 degree-of-freedom and 92 muscle actuators was scaled to data from a static trial for each participant (Delp et al.,). Scaling created a subject-specific model and considered each participant's individual geometry (e.g., segment size, segment mass, or muscle lengths) (Figure 1(a)). Since the dynamic muscle forces of the generic model were too low to perform the dynamic tasks in the current study, the maximum isometric muscle forces of each participant were scaled by a generic (×3) and a subject-specific constant that was based on estimates of lower extremity muscle volume (Handsfield et al.,). Lower extremity muscle volumes were estimated based on regression model (Equation 1) (Handsfield et al.,).

(Equation 1)

$$Volume\ of\ lower\ extremity\ muscles = (47 \times Bodymass \times Height) + 1285$$

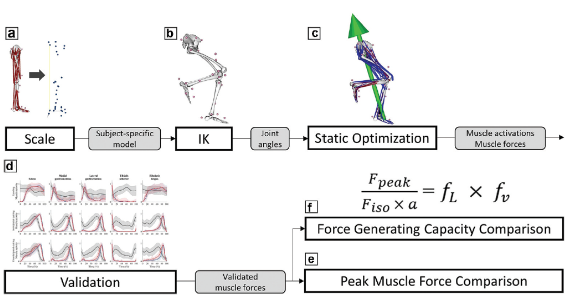


Figure 1. Workflow. (IK: inverse kinematics, $F_{peak}$: peak muscle force, $F_{iso}$: maximum isometric force, a: activation, $f_L$: effect of muscle length, $f_v$: effect of muscle velocity).

The inverse kinematics (IK) tool was used to calculate joint angles by minimising differences between virtual model markers and experimental markers (Figure 1(b)). Static optimisation (SO) was used to estimate muscle forces and activations from GRF and joint angle data and through minimisation of the sum of squared activations of all muscles (Figure 1(c)). All analyses with a musculoskeletal model were performed with OpenSim (Delp et al.,).

Muscle forces from the soleus, medial gastrocnemius, lateral gastrocnemius, tibialis posterior, tibialis anterior, fibularis longus, fibularis brevis, vastus lateralis/medialis/intermedius (grouped together), rectus femoris, superior/middle/inferior fibres of gluteus maximus (grouped together), anterior/middle/posterior fibres of gluteus medius (grouped together), biceps femoris long and short heads/semimembranosus/semitendinosus (grouped together) were calculated and used for statistical analyses. Peak muscle forces from each trial were extracted and normalised by each participant's body weight (BW) (Figure 1(e)). In addition, the force-generating capacity of each muscle group was calculated by dividing peak muscle force $(F_{peak})$ by the maximum isometric force  and concurrent activation $(F_{iso})$ Figure 1(f)), which also accounts for the effects of muscle length $(f_L)$ and velocity $(f_v)$ (Equation 2) (Arnold et al.,). A greater force-generating capacity indicates that a muscle requires less activation to produce the same amount of muscle force.

(Equation 2)

$$Force\ generation\ capacity\ = \frac{F_{peak}}{F_{iso}\ \times\ a} = f_L\ \times\ f_v$$

EMG data were used to validate the simulated muscle activity from static optimisation (Hicks et al.,). Given the absence of maximum voluntary isometric contractions for all muscles, the experimental and simulated EMG data were normalised to the peak value during the dynamic trials and visually compared based on the temporal pattern of muscle activity (Figure 1(d); Hamner & Delp, Hamner et al.,).

Separate two-way analyses of variance (ANOVA) were used to compare peak muscle forces and force-generating capacity of each muscle and muscle group. The independent variables were group (CAI and CON) and task (LAND, ANT, and UNANT). Significant interaction or main effects were followed by Fisher's least significant difference (LSD) procedure to examine pair-wise differences during post-hoc testing. The alpha level for each ANOVA was set to 0.05. Omega-Squared ($\omega^2$) effect-sizes were also calculated (Equation 3).

(Equation 3)

$$\omega^2\ = \frac{SS_{factor}\ -\ df_{factor}\ \cdot\ MS_{error}}{SS_{total} +\ MS_{error}}$$

The $\omega^2$ was considered as very small if between 0 and 0.01, small if between 0.01 and 0.06, medium if between 0.06 and 0.14, and large if greater than 0.14. All statistical analyses were performed in MATLAB (MathWorks, MA, USA).

# Results

Considering electromechanical delay (Corcos et al.,), the simulated EMG data exhibited similar patterns as the experimental EMG data during all tasks and therefore appear to valid for further analysis and processing (Figure 2; Hicks et al.,).

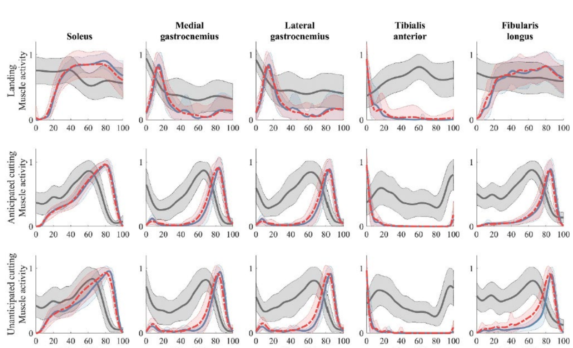

Figure 2. Mean and standard deviation of the simulated muscle activations from static optimisation and experimental EMG (blue line and shaded area: chronic ankle instability group, red line, and shaded area: control group, green shaded area: measured EMG).

There were no significant group by task interactions for any peak muscle forces. However, there was a significant group main effect (p = 0.018) and medium effect size ($\omega^2$= 0.08) for peak gluteus maximus force (Table 2 and Figure 3). Specifically, the CAI group generated greater peak gluteus maximus forces during all tasks.

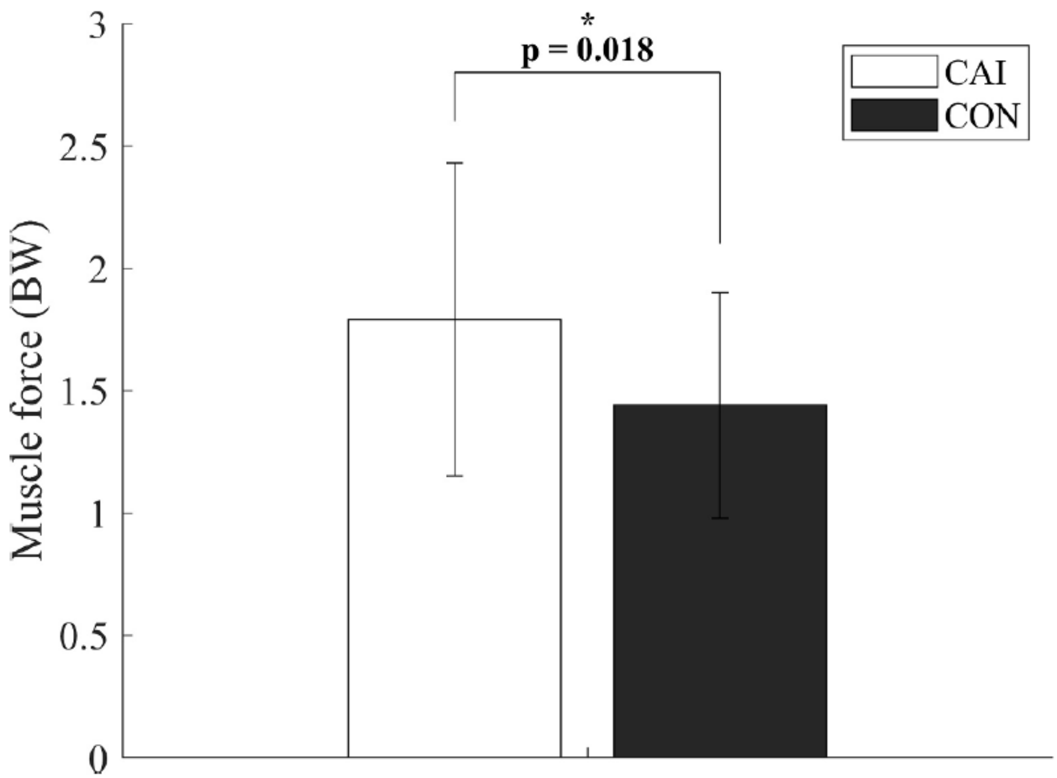


Figure 3. Task-averaged gluteus maximus muscle forces (x body weight) for people with (CAI) and without (CON) chronic ankle instability.

Table 2. Mean and standard deviation of muscle force. (CAI: chronic ankle instability group, CON: control group, LAND: landing, ANT: anticipated cutting, UNANT: unanticipated cutting, SL: soleus, MG, medial gastrocnemius, LG: lateral gastrocnemius, TA: tibialis anterior, FL: fibularis longus, VAS: vastus muscles, RF: rectus femoris, GX: gluteus maximus, GM: gluteus medius, HAMS: hamstrings, †: significant group main effect in two-way ANOVA)

| | | | | Muscle Force (BW) | | | | | Two-way ANOVA | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | CAI | | | CON | | Group | | | Interaction | | Group | | Task | |
| Muscle | LAND | ANT | UNANT | LAND | ANT | UNANT | CAI | CON | p | $\omega^2$ | p | $\omega^2$ | p | $\omega^2$ |
| SL | 4.69±0.78 | 5.50±1.10 | 6.04±1.20 | 4.32±1.23 | 5.79±0.94 | 5.80±0.87 | 5.37±1.14 | 5.30±1.21 | 0.560 | 0.01 | 0.687 | 0.01 | 0.001 | 0.25 |
| MG | 1.20±0.34 | 1.55±0.29 | 1.78±0.53 | 1.05±0.37 | 1.59±0.38 | 1.45±0.28 | 1.49±0.45 | 1.37±0.41 | 0.297 | 0.01 | 0.133 | 0.02 | 0.001 | 0.25 |
| LG | 0.29±0.09 | 0.37±0.11 | 0.47±0.21 | 0.26±0.10 | 0.39±0.15 | 0.37±0.13 | 0.37±0.15 | 0.34±0.14 | 0.353 | 0.00 | 0.308 | 0.00 | 0.003 | 0.16 |
| TA | 0.20±0.10 | 0.18±0.07 | 0.24±0.11 | 0.28±0.19 | 0.14±0.06 | 0.26±0.13 | 0.20±0.10 | 0.23±0.15 | 0.335 | 0.00 | 0.464 | 0.01 | 0.025 | 0.09 |
| FL | 0.14±0.18 | 0.76±0.33 | 0.95±0.49 | 0.10±0.07 | 0.85±0.52 | 0.60±0.27 | 0.59±0.48 | 0.52±0.46 | 0.136 | 0.02 | 0.268 | 0.00 | 0.001 | 0.44 |
| VAS | 6.98±0.88 | 7.49±0.76 | 8.13±1.43 | 6.66±1.41 | 8.20±1.82 | 8.42±2.11 | 7.49±1.10 | 7.76±1.91 | 0.522 | 0.01 | 0.543 | 0.01 | 0.008 | 0.13 |
| RF | 1.35±0.41 | 1.11±0.53 | 1.87±0.97 | 1.70±0.64 | 1.28±0.24 | 1.58±0.44 | 1.42±0.70 | 1.52±0.49 | 0.201 | 0.02 | 0.605 | 0.01 | 0.016 | 0.10 |
| GX | 1.77±0.64 | 1.70±0.61 | 1.92±0.73 | 1.33±0.38 | 1.46±0.49 | 1.54±0.52 | 1.79±0.64† | 1.44±0.46 | 0.843 | 0.03 | 0.018 | 0.08 | 0.579 | 0.01 |
| GM | 3.31±0.53 | 2.48±0.69 | 3.09±1.04 | 3.19±0.51 | 2.39±0.60 | 2.90±0.87 | 2.95±0.82 | 2.83±0.73 | 0.976 | 0.03 | 0.479 | 0.01 | 0.002 | 0.17 |
| HAMS | 1.24±0.39 | 1.68±0.30 | 1.99±0.69 | 1.16±0.27 | 1.53±0.54 | 1.91±0.54 | 1.61±0.55 | 1.53±0.55 | 0.963 | 0.02 | 0.394 | 0.00 | 0.001 | 0.29 |

There was a significant group by task interaction (p = 0.009) with medium effect size ($\omega^2$= 0.11) for force-generating capacity of vastii (Table 3). Fisher's post hoc test revealed that force-generating capacity of vastii was significantly greater in CAI group compared to the CON group during UNANT (p = 0.001) (Table 3 and Figure 5). Furthermore, there was a significant group main effect (p = 0.021) with medium effect size ($\omega^2$= 0.06) for force-generating capacity of gluteus maximus (Table 3 and Figure 4). Specifically, force-generating capacity of gluteus maximus was significantly greater in CAI group compared to the CON group regardless of task.

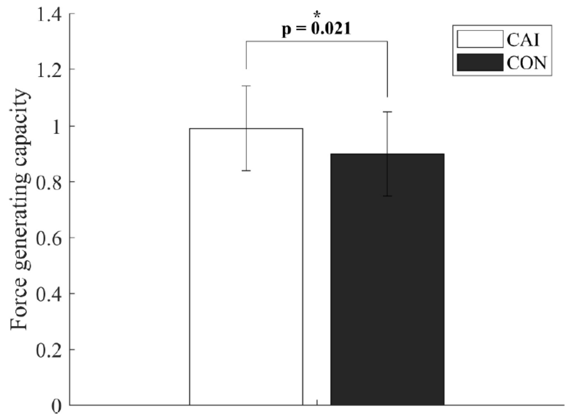


Figure 4. Task-averaged gluteus maximus force generating capacity muscle forces for people with (CAI) and without (CON) chronic ankle instability.

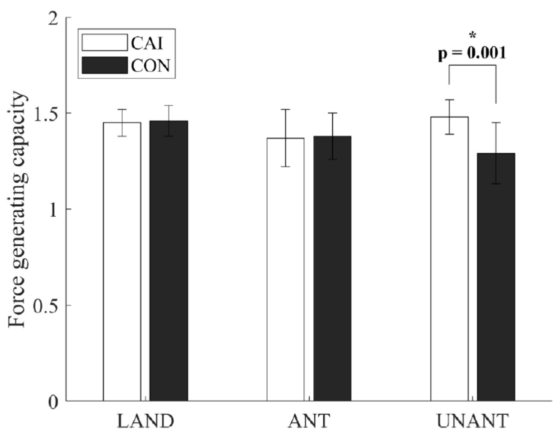


Figure 5. Vastii muscle group force generating capacity for people with (CAI) and without (CON) chronic ankle instability during the landing (LAND), anticipated cutting (ANT), and unanticipated cutting (UNANT) tasks.

Table 3. Mean and standard deviation of force generating capacity. (CAI: chronic ankle instability group, CON: control group, LAND: landing, ANT: anticipated cutting, UNANT: unanticipated cutting, SL: soleus, MG, medial gastrocnemius, LG: lateral gastrocnemius, TA: tibialis anterior, FL: fibularis longus, VAS: vastus muscles, RF: rectus femoris, GX: gluteus maximus, GM: gluteus medius, HAMS: hamstrings, †: significant group main effect in two-way ANOVA, ‡, §, ¶, #: significant differences in Fisher's LSD post hoc testing)

| | | | | Force generating capacity | | | | | Two-way ANOVA | | | | | |
|---|---|---|---|---|---|---|---|---|---|---|---|---|---|---|
| | CAI | | | CON | | | Group | | Interaction | | Group | | Task | |
| Muscle | LAND | ANT | UNANT | LAND | ANT | UNANT | CAI | CON | p | $\omega^2$ | p | $\omega^2$ | p | $\omega^2$ |
| SL | 1.34±0.08 | 0.95±0.25 | 1.08±0.15 | 1.36±0.12 | 0.92±0.14 | 0.98±0.16 | 1.13±0.24 | 1.08±0.24 | 0.506 | 0.01 | 0.341 | 0.01 | 0.001 | 0.55 |
| MG | 1.29±0.11 | 0.52±0.28 | 0.77±0.40 | 1.20±0.14 | 0.49±0.28 | 0.85±0.38 | 0.86±0.43 | 0.85±0.40 | 0.668 | 0.01 | 0.879 | 0.01 | 0.001 | 0.54 |
| LG | 1.38±0.11 | 0.54±0.33 | 0.80±0.47 | 1.29±0.15 | 0.48±0.32 | 0.95±0.45 | 0.91±0.48 | 0.91±0.46 | 0.492 | 0.01 | 0.994 | 0.01 | 0.001 | 0.52 |
| TA | 0.55±0.07 | 0.66±0.18 | 0.53±0.07 | 0.62±0.12 | 0.68±0.17 | 0.55±0.12 | 0.59±0.13 | 0.62±0.14 | 0.765 | 0.02 | 0.269 | 0.01 | 0.008 | 0.13 |
| FL | 1.26±0.17 | 0.61±0.25 | 0.62±0.18 | 1.34±0.18 | 0.54±0.05 | 0.69±0.17‡§# | 0.84±0.37 | 0.86±0.38 | 0.348 | 0.01 | 0.536 | 0.01 | 0.001 | 0.78 |
| VAS | 1.45±0.07‡ | 1.37±0.15 | 1.48±0.09# | 1.46±0.08§ | 1.38±0.12 | 1.29±0.16 | 1.43±0.12 | 1.38±0.14 | 0.009 | 0.11 | 0.061 | 0.04 | 0.080 | 0.04 |
| RF | 1.46±0.18 | 1.25±0.34 | 1.50±0.16 | 1.34±0.19 | 1.30±0.27 | 1.49±0.13 | 1.40±0.26 | 1.38±0.21 | 0.504 | 0.01 | 0.672 | 0.01 | 0.012 | 0.11 |
| GX | 1.08±0.18 | 0.91±0.07 | 0.97±0.14 | 0.95±0.16 | 0.84±0.15 | 0.90±0.14 | 0.99±0.15† | 0.90±0.15† | 0.768 | 0.02 | 0.021 | 0.06 | 0.011 | 0.11 |
| GM | 1.28±0.12 | 1.08±0.17 | 1.11±0.18 | 1.25±0.07 | 1.01±0.10 | 1.06±0.12 | 1.16±0.17 | 1.11±0.14 | 0.846 | 0.02 | 0.167 | 0.01 | 0.001 | 0.34 |
| HAMS | 0.72±0.09 | 0.88±0.10 | 0.81±0.12 | 0.69±0.06 | 0.82±0.15 | 0.78±0.07 | 0.80±0.12 | 0.76±0.11 | 0.859 | 0.02 | 0.110 | 0.02 | 0.001 | 0.23 |

## Discussion and implications

The purpose of the current study was to estimate peak forces and force-generating capacity of lower extremity muscles and to compare these estimates between a group of people with CAI and a healthy control group as they both performed landing and cutting tasks. The results showed that people with CAI exhibited greater peak gluteus maximus muscle forces and a greater capacity to generate gluteus maximus force than people in the CON group across all tasks. In addition, the CAI group also exhibited greater vastii force-generating capacity than the CON group during the unanticipated cutting task. Together, these results partially supported our hypotheses that people with CAI would exhibit different peak muscle forces and force-generating capacity, and that these differences would depend on the respective task.

A primary finding of the current study was that people with CAI generated greater peak gluteus maximus forces than people in the CON group across all tasks. More specifically, people with CAI generated on average approximately 24% greater peak gluteus maximus forces during all landing and cutting tasks. This finding agrees with previous studies, which reported that people with CAI exhibit compensatory muscle activations at proximal joints (DeJong et al., K. Kim et al., Rios et al.,). For example, H. Kim et al. observed greater activations of knee and hip joint muscles (e.g., vastus lateralis, adductor longus, gluteus maximus, and gluteus medius) in CAI patients during the transition phase (i.e., after landing and before takeoff) of landing/cutting tasks (K. Kim et al.,). Similarly, Rios et al. reported that people with CAI activated muscles at proximal joints more during the single-leg stance phase of ball-kicking tasks than a group of healthy controls (Rios et al.,). Furthermore, DeJong et al. observed that the difference in ultrasound-based gluteus maximus muscle thickness, which is a purported surrogate of muscle activation, between resting and exercise conditions during a dynamic balance task were greater in a group of people with CAI than a group of healthy controls (DeJong et al.,]; Mangum et al., ). The authors of these studies suggested that people with CAI adopt greater activation of proximal muscles as a compensatory strategy that aims to improve postural control and mitigate neuromuscular deficits at the ankle joint (DeJong et al., K. Kim et al., Rios et al.,). However, since muscle activation assessed via EMG or ultrasound only provide indirect, and somewhat tenuous, information about muscle forces, the current study provides more direct evidence that compensatory muscle function in people with CAI also extends to the generation of force in proximal muscles. Specifically, the greater gluteus maximus force in the current study bolsters previous assertions that people with CAI stabilise proximal joints and segments across the kinetic chain to help stabilise distal joints (Webster et al.,). This interpretation is supported by the fact that the gluteus maximus has an important role during the performance of athletic tasks because it prevents excessive hip adduction, trunk flexion angle, and femoral internal rotation angles, which may deleteriously affect ankle inversion through kinematic coupling (Buckthorpe et al., MacKinnon & Winter,). Collectively, these findings therefore suggest that people with CAI exhibit neuromuscular differences in the function of proximal muscles, which may reflect a strategy to compensate for deficits at the ankle joint. Despite these findings, further studies should investigate association between proximal muscle forces and ankle joint kinematics to provide direct evidence about the compensatory mechanism.

Another finding of the current study was that people with CAI exhibited an approximately 10% greater gluteus maximus force-generating capacity during all tasks. Given that muscle's force generating capacity results from the interaction of the force-length $(f_L)$ and force-velocity $(f_v)$ behaviour that it exhibits during dynamic tasks, the above result indicates that people with CAI performed all landing and cutting tasks with the gluteus maximus operating closer to its optimal length and/or with slower shortening velocities than people in the CON group. In addition, this result suggests that people with CAI may produce gluteus maximus force more efficiently than people in the CON group since operating at a greater force-generating capacity could also reflect that a given muscle requires less activation to generate a specific amount of force. It is thus likely that people with CAI generated greater gluteus maximus force because their chosen movement strategy allowed them to operate at a greater force-generating capacity. It is also interesting to note that people with CAI exhibited an approximately 15% greater force-generating capacity of vastii, but only during the UNANT task. In contrast to the results about group differences in the gluteus maximus force-generating capacity, the difference in force-generating capacity of the vastii muscles between groups therefore appears to be task-dependent. The task-dependent difference in force-generating capacity likely indicates that during unanticipated tasks, people with CAI alter their movement strategy to allow them to operate at a greater force-generating capacity of the vastii, in addition to that of the gluteus maximus, in order to compensate for the uncertainty that is inherent in this task. Since the differences in vastii force-generating capacity were not accompanied by differences in peak muscle forces, one could speculate that people with CAI exhibit greater efficiency in generating vastii muscle forces during tasks with uncertainty, which may reflect a novel aspect of their compensatory strategy that has not been previously characterised.

We acknowledge several limitations and provide considerations for future study. First, the markers that were used to define the foot segment in the current study were attached to the outside of the participant's shoes, and it is acknowledged that movement of these markers may not directly represent movement of the foot segment. Second, we assumed that the foot segment is one rigid body in this study. However, single-segment foot models may not adequately represent the exact kinematics of the ankle joint (K. Kim & Kipp). The use of multi-segment foot model can be used for better-capturing foot and ankle movement for the future study. Third, the validation of simulated muscle activations involved only five ankle muscles and did not include proximal muscles. However, the cost function of the SO algorithm minimises the sum of squared activations of all muscles (proximal and distal), and all estimates fell within reasonable ranges. Although previous research suggests that results from the SO algorithm provide a better match with experimental data than other algorithms (Karabulut et al.,), future studies should consider collecting EMG from other muscles for a more comprehensive validation of simulated results.

## Conclusions

The current study bolsters the evidence of neuromuscular deficits and task-specific compensatory movement strategies in CAI patients during dynamic movements. Compensatory movement strategies in the CAI patients, such as a 'proximal dominant landing strategy', have been reported in previous studies based on EMG or joint kinematic/kinetic findings (K. Kim et al.,). The current findings provide additional evidence that previously observed disparities in muscle function also extend to differences in muscle forces and force generation capacities (i.e., force-length-velocity behaviour) of proximal muscles, and should thus also be considered part of a compensatory landing strategy, in people with CAI. Because compensatory function of proximal muscles suggested that people with CAI exhibited greater muscles force and larger force generation capacities, future research about CAI rehabilitation interventions that aim to prevent recurrent injuries should consider training proximal movement patterns (e.g., kinematic profile) rather than only strengthening muscles, especially during dynamic tasks that include unanticipated decision-making elements.


## Disclosure statement

No potential conflict of interest was reported by the author(s).


## Footnotes

1 *The research presented in this study was conducted at University of Michigan*

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