



Original Research

Effects of foot orthoses on walking and jump landing biomechanics of individuals with chronic ankle instability

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ABSTRACT

Objectives: To evaluate the kinematic, kinetic and electromyographic (EMG) immediate effects of foot orthoses (FOs) during walking, maximal single-leg side jump landing (SIDE) and unilateral drop landing on even (DROP), inclined and unstable (FOAM) surfaces in individuals with chronic ankle instability (CAI).

Design: Cohort study.

Setting: Biomechanics laboratory.

Participants: 26 healthy individuals with CAI.

Main outcome measures: Ankle and knee angles/moments and lower-limb EMG of the gluteus medius, vastus medialis, vastus lateralis, biceps femoris, gastrocnemius medialis, gastrocnemius lateralis, peroneus longus and tibialis anterior muscles.

Results: The main results are that with FOs, individuals with CAI exhibited decreased tibialis anterior muscle activity from 19 to 38% and 39–99% of the landing phase during the DROP task. They also exhibited increased biceps femoris muscle activity from 56 to 65% of the preactivation phase during walking. No significant ankle and knee joints angles and moments difference was observed when wearing FOs in any of the experimental tasks.

Conclusions: The results of this study suggest that the biomechanical effects of FOs are task-dependent and only affect EMG activity. They will help clinicians and researchers to better understand FOs' role in treatment and prevention of CAI.

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1. Introduction

Ankle sprains are very common during sports, representing 10–30% of all musculoskeletal injuries (Fong, Hong, Chan, Yung, & Chan, 2007) but can also occur during daily activities (Gribble et al., 2016). As much as 32–74% of individuals sustaining an ankle sprain will develop chronic ankle instability (CAI) (Hiller et al., 2012; Tanen, Docherty, Van Der Pol, Simon, & Schrader, 2014). CAI alters lower limb biomechanics during dynamic tasks and these deficits could place the individuals more at risk of sustaining recurrent ankle sprains (Moisan, Descarreaux, & Cantin, 2017; Simpson, Stewart, Macias, Chander, & Knight, 2019). Individuals with CAI

show biomechanical deficits during daily activities, such as walking (Moisan et al., 2017). They exhibit increased peroneus longus muscle activity (Delahunt, Monaghan, & Caulfield, 2006a), ankle joint inversion (Delahunt et al., 2006a) and a laterally deviated centre-of-pressure (Koldenhoven, Feger, Fraser, Saliba, & Hertel, 2016) compared to healthy controls.

In previous studies, unilateral jump landing tasks have also been commonly used to assess the biomechanical deficits associated with CAI (Simpson et al., 2019), mostly because ankle sprains are very common during activities involving unilateral jumping and landing such as basketball, volleyball and soccer (Yeung, Chan, So, & Yuan, 1994). The high loads put on the ankle joint complex during these movements have been reported as a risk factor to sustain a lateral ankle sprain (Doherty et al., 2016). During unilateral drop landing, individuals with CAI demonstrate greater ankle dorsiflexion pre and post-landing (Caulfield & Garrett, 2002),

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greater ankle inversion pre-landing (Delahunt, Monaghan, & Caulfield, 2006b), greater knee flexion pre and post-landing (Caulfield & Garrett, 2002), altered ground reaction forces (Delahunt et al., 2006b) and decreased peroneus longus muscle preactivation (Delahunt et al., 2006b). Alterations of foot and ankle kinematics have also been observed during a maximal side jump landing (De Ridder et al., 2015). The effects of CAI on ankle and knee moments during unilateral jump landing is still not well understood (Simpson et al., 2019).

Previous studies quantified the effects of CAI during unilateral jump landing on an even surface. However, these tasks only partially represent the situations during which individuals sprain their ankle on a variety of landing surfaces (e.g. uneven, irregular or unstable). Biomechanical deficits associated with CAI have been observed during more challenging tasks such as a drop landing on a 20-degree laterally inclined surface (Simpson et al., 2019). Individuals with CAI exhibit prolonged peroneus longus muscle latency, increased maximal ankle inversion and decreased time to maximal ankle inversion. Landing on an unstable surface such as a foam pad also greatly reduces lower limb muscles preactivation (Prieske et al., 2013). The aforementioned deficits could place a CAI ankle at a higher risk of giving way or sprain, especially when landing on inclined or unstable surfaces. Assessing these tasks will improve our knowledge of the biomechanical deficits associated with CAI during more challenging situations.

Foot orthoses (FOs) have been extensively used in previous studies as a treatment for musculoskeletal pathologies mainly because they can modulate lower limb kinematics (Telfer, Abbott, Steultjens, & Woodburn, 2013), kinetics (Telfer et al., 2013) and muscle activity (Moisan & Cantin, 2016) during dynamic tasks, such as walking and jump landing. Even though no study has yet quantified the biomechanical effects of FOs in individuals with CAI during walking and jump landing, FOs can influence neuromuscular control during a transition from double to single-legged stance for this population (Dingenen et al., 2015). It has been hypothesised that sensory information coming from the foot may become more reliable when wearing FOs because the conformity of the longitudinal arch with the foot may provide enhanced plantar cutaneous sensation, increase the efferent activity and thus enhance neuromuscular control (Sesma, Mattacola, Uhl, Nitz, & McKeon, 2008). FOs could perhaps optimise the efficacy of current treatment modalities, such as neuromuscular training, which often fail to completely restore the neuromuscular deficits associated with CAI (Doherty, Bleakley, Delahunt, & Holden, 2017). Measuring the FOs' effects during walking and jump landing tasks will help to better target the biomechanical deficits associated with CAI that can be attenuated and will provide proof-of-concept and indications if future efficacy trials are warranted.

The primary objective of this study is to quantify the kinematic, kinetic and electromyographic (EMG) immediate effects of FOs during walking and unilateral jump landing in individuals with CAI. The main hypothesis of this study is that FOs will decrease the peroneus longus and tibialis anterior muscles activity and the ankle joint range of motion and inversion moment during locomotion. It was also hypothesised that FOs' effects would be greater during more challenging tasks.

2. Materials and methods

2.1. Participants

An a priori sample size calculation was performed with G-Power (Version 3.1, Kiel, Germany) software based on the results of a previous study that investigated ankle sagittal kinematics during walking (Dedieu, Drigeard, Gjini, Dal Maso, & Zanone, 2013).

Considering $\alpha = 0.05$, $1 - \beta = 0.80$ and Cohen's d effect size of 0.59, a sample size of 26 participants was chosen. The walking task was used for the sample size calculation, as it is the task that required the greatest number of participants. The 26 participants (Age: 25.3 ± 5.2 years, weight: 72.2 ± 13.2 kg, height: 1.69 ± 0.10 m, number of sustained sprains: 4.5 ± 3.9 , time since last sprain: 1.8 ± 1.9 years, Foot Posture Index (FPI) score: 4.7 ± 3.3 , FAAM-ADL: 83.8 ± 8.5 , FAAM-S: 61.2 ± 10.7 and IPAQ: 3421 ± 3215 MET-min/week) were recruited among the UQTR students and staff, from the university's outpatient podiatry clinic and through advertisement on social media between June and December 2018. The inclusion criteria, based on the International Ankle Consortium recommendations (Gribble et al., 2014), were (1) a score of respectively $<90\%$ and $<80\%$ for the Foot and Ankle Ability Measure Activity of daily living (FAAM-ADL) and Sport (FAAM-S) subscales, (2) a history of at least one significant ankle sprain that occurred one year or more prior to the study onset and (3) a history of ankle joint giving way and/or having recurrent ankle sprains and/or perceive the ankle as unstable. The limb with the less stable ankle, chosen by the participants, was evaluated when they had bilateral CAI. The exclusion criteria were (1) a history of surgery to the musculoskeletal structures or fracture of the lower extremity that required surgical realignment, (2) musculoskeletal injury to the lower extremity and having worn FOs on a regular basis in the 3-month period prior to the study onset and (3) any known condition that affect the participants' ability to walk or jump. Prior to their participation, all participants gave their written informed consent according to the protocol ethics certification approved by the university ethics committee (CER-18-244-07.04).

2.2. Experimental protocol

Two weeks prior to the experimental session, a licensed podiatrist took non-weightbearing plaster casts of the participants' feet with the subtalar joint in neutral position and the midtarsal joint in maximal pronation as described by Root et al. (Root, Weed, & Orien, 1971). The casts were balanced to vertical and a minimal medial arch fill was added. The FOs used in this study were made by a certified orthotist technician from 3.2 mm thick polypropylene with a straight ethylene-vinyl-acetate (EVA) rearfoot post, a EVA lateral bar (Moisan & Cantin, 2016; Moisan, Descarreaux, & Cantin, 2018) and a 3 mm multiform full-length top cover. The participants also had to fill validated French version of the FAAM-ADL, FAAM-S (Borloz et al., 2011) and the International Physical Activity Questionnaire (IPAQ) (Criniere et al., 2011). They also reported the number of previously sustained ankle sprains and the time since the most recent sprain. To quantify the participants' foot morphology, the FPI was used (Redmond, Crosbie, & Ouvrier, 2006).

During the experimental session, a wireless surface EMG system (Delsys Trigno, Boston, USA) was used to record muscle activity of the gluteus medius, vastus lateralis, vastus medialis, biceps femoris, gastrocnemius lateralis, gastrocnemius medialis, peroneus longus and tibialis anterior. To reduce local impedance, the skin was shaved, gently abraded with fine-grade sandpaper and swiped with alcohol swabs. Electrode positioning was carried out according to the SENIAM recommendations (Hermens, Freriks, Disselhorst-Klug, & Rau, 2000). The 99% silver contact material electrodes with a 4-bar formation had a dimension of $27 \text{ mm} \times 37 \text{ mm} \times 15 \text{ mm}$. The inter-electrode distance was 10 mm, the common noise removal ratio of the amplifier was $>80 \text{ dB}$ and a 16 bits A/D converter was used. EMG data were sampled at 2000 Hz, the gain was 1000 and the maximum inraelectrode impedance was 6 kOhm.

Kinematic data were collected with an active three-dimensional motion analysis system (Optotrak Certus, Northern Digital, Ontario, Canada) at a sampling frequency of 100 Hz. Four three-marker rigid

plates were used to capture the lower extremity kinematics. These plates were positioned on the sacrum, the distal one third of the thigh, the distal one third of the leg and the posterior part of the calcaneum using a modified version of the wand (manufactured to be thicker and less brittle) described by Telfer et al. (Telfer et al., 2013). The calcaneal wand was attached to the posterior portion of the calcaneum with a customised plastic heel plate secured to the participants' skin with athletic tape. To allow the calcaneal wand to be attached directly onto the foot, a rectangular hole of approximately 30 mm × 30 mm was cut in the shoe heel counter (Athletic Works, Model: Rupert) over a standardised location. During a static trial, 15 virtual kinematic markers were created on the pelvis and the tested lower extremity with a digitizing pointer: bilateral anterior superior iliac spines, bilateral posterior superior iliac spines, greater trochanter, lateral and medial femoral condyles, fibular head, tibial tuberosity, medial and lateral malleoli, proximal posterior surface of calcaneus, distal attachment of the Achilles' tendon, sustentaculum tali and fibular tubercle. These markers were used to create the pelvis, hip, knee, ankle and foot segments, precisely locate the hip/knee/ankle joint centres and finally calculate ankle and knee 3D movements and moments.

Ground reaction forces data, sampled at 2000Hz, were collected with a force plate (Bertec Corp, OH, USA) embedded in the floor. To measure walking speed, electronic photocells timing gates (Brower Timing System, USA) positioned 1.53 m before and after the force plate were used. A 3.8 cm × 3.8 cm footswitch was positioned in the shoe, under the heel of the tested limb (Trigno 4-Channel FSR Adapter, Boston, USA).

All participants completed five trials of five tasks with and without FOs in a randomly decided order across participants for a total of 50 trials (25 with FOs, 25 shod). Before the 25 trials of both conditions (FOs and shod), a static trial was performed. The first task consisted of walking at a self-selected speed on a 7.5-m walkway with an embedded force plate positioned halfway on the participants' path. First, five familiarisation trials during which walking speed was recorded and averaged were completed. Then, the trials used in the analysis were completed using a midgait protocol. All trials during which speed varied more than 5% of the mean walking speed established during the familiarisation trials or the foot did not entirely hit the force plate were rejected and immediately retaken. The second task was a maximal single-leg side jump (SIDE). Participants started the task with only the tested foot on the floor. They had to laterally jump as far as possible, land with the same foot on a force plate and stabilise for at least 2 s. The third task was a single-leg drop jump (DROP) from a 46 cm high platform. The participants had to anteriorly propel themselves with the contralateral leg, land on the centre of the force plate with the tested leg and stabilise for 2 s. They were instructed to carefully drop off the platform without an upward jump action in order to standardise the drop height. The fourth task was a single-leg drop jump during which they had to land on a 10-cm thick foam block (FOAM) positioned over the force plate. The fifth task was a single-leg drop jump during which the participants had to land with the tested foot on a 25° laterally inclined wood surface (WEDGE) positioned over the force plate. The platform height was adjusted during the FOAM and WEDGE tasks for the overall drop distance to be 46 cm. During all jump landing tasks, the participants were asked to place both hands on their waist, face and look forward. The participants were allowed to practice jump landing until they were comfortable performing the tasks.

2.3. Data processing

EMG, kinematic and kinetic data were processed using Visual3D software (C-motion, Inc., Germantown, MD, USA). EMG data were

digitally filtered with a zero phase lag, bi-directional, 20–450 Hz bandpass fourth-order Butterworth filter. Analyses were performed on the Root Mean Square (RMS) of these data, calculated with a 100 ms moving window. RMS data of each muscle (shod and FOs conditions) were normalised with the mean peak RMS amplitude of all trials of the shod condition for each individual task. The EMG data were not normalised with maximal voluntary isometric contractions, as it elicited pain for some participants. The submaximal normalisation technique was previously used in studies investigating FO's EMG effects during locomotion (Moisan & Cantin, 2016, 2017; Moisan et al., 2018). Kinematic data were low-pass filtered at 6 Hz by a dual-pass, fourth-order Butterworth filter. Ankle and knee joint angles were calculated using a Cardan sequence with order of X (extension/flexion), Y (adduction/abduction), and Z (internal/external rotation). The force plate data were low-pass filtered by a dual pass, fourth-order Butterworth filter with a cut-off frequency of 50 Hz and were used to calculate ankle and knee joints moments by inverse dynamics. Joint angles and moments were resolved in the proximal segment coordinate system. Joint moments were normalised to body mass. Joint moments and angles were normalised to 100% of the landing phase for each task. The EMG data were also normalised to 100% of the preactivation phase. For the walking task, 0% of the landing phase represented the initial foot contact with the force plate and 100% represented the toe off. For the preactivation phase, the toe off and initial foot contact with the force plate were respectively defined as 0% and 100%. For the jump landing tasks, 0% of the landing phase represented the initial foot contact with the force plate or the surfaces (FOAM and WEDGE) and 100% represented the maximal knee flexion. For the preactivation phase, the heel off and the initial foot contact with the force plate or the surfaces (FOAM and WEDGE) were respectively defined as 0% and 100%. A threshold set at 10 N was used to determine the task event with the force plate.

2.4. Statistical analysis

The normality of walking speed data was assessed with the Shapiro-Wilk test. As the data were not normally distributed, the Wilcoxon test was used to compare data between conditions with a level of significance of $p < 0.05$. As the kinematic, kinetic and EMG data were not all normally distributed according to the D'Agostino-Pearson test, the one-dimensional non-parametric permutation method (SnPM) was used to compare each individual point of the curves between both experimental conditions with a threshold of $\alpha = 5\%$ (Nichols & Holmes, 2002). The individual probability that each supra-threshold cluster could have resulted from an equivalently smooth random process was determined. Cohen's d effect sizes were calculated when statistically significant differences were observed. All SnPM analyses were implemented using the open-access SPM1D code (www.spm1d.org) in Python software (Version 2.7).

3. Results

No statistically significant difference in walking speed between shod and FOs conditions (1.41 ± 0.15 vs 1.41 ± 0.14 m/s, $p = 0.39$) was observed.

For the DROP task, a decreased tibialis anterior muscle activity was observed with FOs from 19 to 38% ($p = 0.01$) and 39–99% ($p < 0.01$) of the landing phase (see Fig. 1a). The gastrocnemius medialis muscle activity was increased from 11 to 18% ($p < 0.01$) of the preactivation phase when wearing FOs (see Fig. 1b). For the FOAM task, an increased gastrocnemius lateralis muscle activity was observed with FOs from 16 to 17% ($p = 0.02$) and 18–26% ($p < 0.01$) of the preactivation phase (see Fig. 2). For the walking

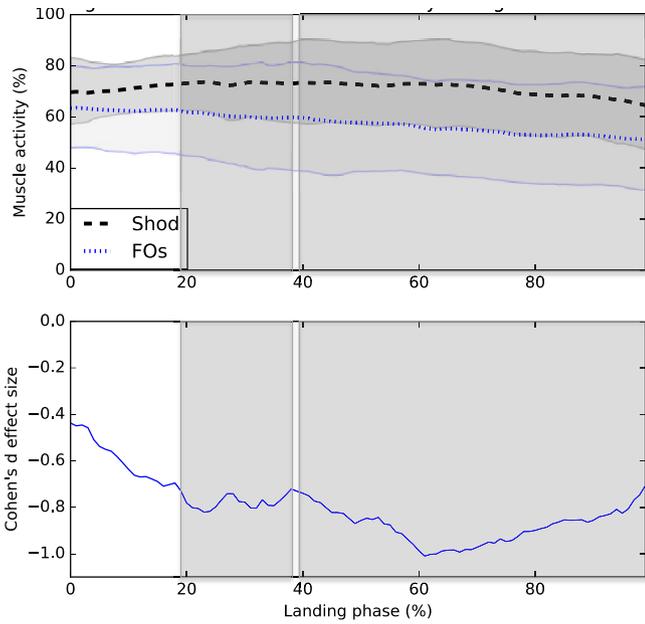


Fig. 1a. Tibialis anterior muscle activity during the DROP task.

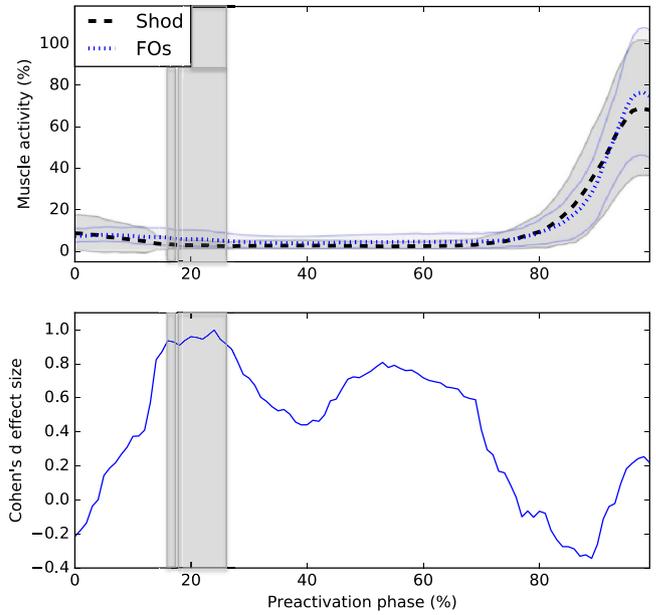


Fig. 2. Gastrocnemius lateralis muscle activity during the FOAM task.

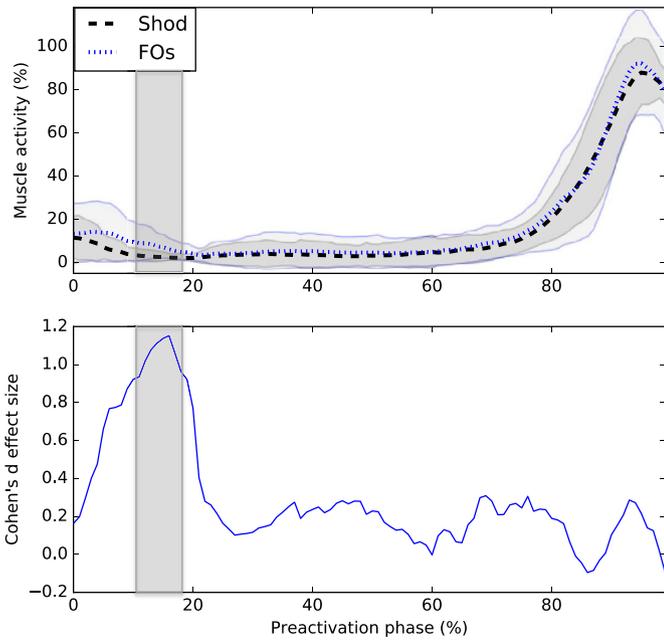


Fig. 1b. Gastrocnemius medialis muscle activity during the DROP task.

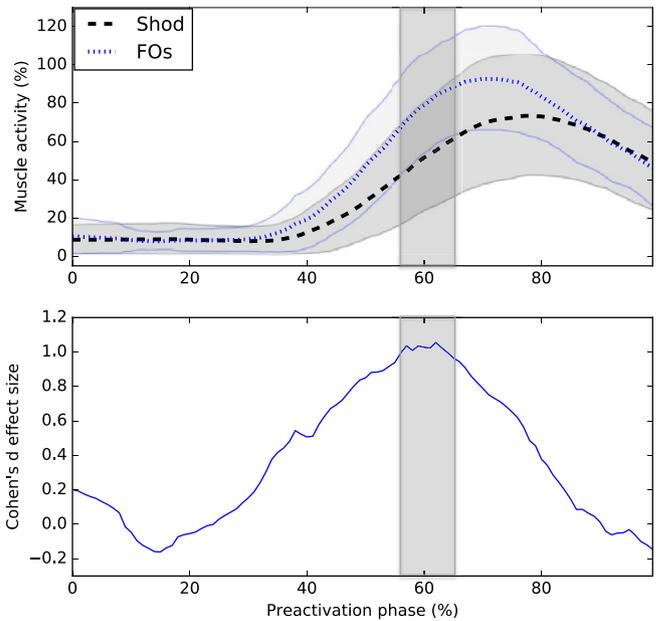


Fig. 3. Biceps femoris muscle activity during walking.

task, an increased biceps femoris muscle activity was observed with FOs from 56 to 65% ($p = 0.01$) of the preactivation phase (see Fig. 3). No significant difference was observed during the WEDGE and SIDE tasks with shoes and with FOs for EMG, kinematics and kinetics. No significant difference was observed for ankle and knee angles and moments during any of the experimental tasks. All kinematic, kinetic and EMG dataset are available in supplementary files.

Figure captions: The upper panel shows the mean and standard deviation of the tibialis anterior muscle EMG activity measured during the landing phase of the DROP task with and without FOs. The lower panel shows the Cohen's d effect size during the same period. Significant effects of FOs occurred during the period

indicated by the shadowed regions.

Figure captions: The upper panel shows the mean and standard deviation of the gastrocnemius medialis muscle EMG activity measured during the preactivation phase of the DROP task with and without FOs. The lower panel shows the Cohen's d effect size during the same period. Significant effects of FOs occurred during the period indicated by the shadowed regions.

Figure captions: The upper panel shows the mean and standard deviation of the gastrocnemius lateralis muscle EMG activity measured during the preactivation phase of the FOAM task with and without FOs. The lower panel shows the Cohen's d effect size during the same period. Significant effects of FOs occurred during the period indicated by the shadowed regions.

Figure captions: The upper panel shows the mean and standard deviation of the biceps femoris muscle EMG activity measured during the preactivation phase of the walking task with and without FOs. The lower panel shows the Cohen's *d* effect size during the same period. Significant effects of FOs occurred during the period indicated by the shadowed regions.

4. Discussion

The main finding of this study was that individuals with CAI exhibited decreased tibialis anterior muscle activity from 19 to 38% and 39–99% of the landing phase during the DROP task, with Cohen's *d* effect sizes ranging from -0.73 to -1.01 . A recent systematic review hypothesised that individuals with CAI present altered ankle biomechanics, such as increased ankle dorsiflexion, in order to place the talo-crural joint in a tightly packed position to increase its stability (Simpson et al., 2019). Even though FOs had no effect on ankle joint angles during the DROP task, the decreased tibialis anterior muscle activity could represent increased ankle stability when wearing FOs. No comparison could be made with previous studies, as this is the first study to quantify the FOs' EMG effects for individuals with CAI during a DROP task. However, even though the FOs' effects on mean tibialis anterior muscle activity were not investigated, earlier activation was observed when individuals with CAI wore FOs during a transition from double-legged to single-legged stance (Dingenen et al., 2015). Even though further studies are needed to better understand their effects on tibialis anterior muscle activity during dynamics tasks, the results of the current study and those of Dingenen et al. (2015) suggest a better neuromuscular control when wearing FOs for individuals with CAI.

The second finding of this study is the increased gastrocnemius medialis and gastrocnemius lateralis during the preactivation phase of respectively the DROP and FOAM tasks. The increased activity was observed during the propulsion phase (on the platform) and could suggest alterations of the motor control for individuals with CAI when wearing FOs during the DROP task. However, even though the Cohen's *d* effect sizes are large (0.91–1.00), these results should be interpreted with caution as they were observed during low-velocity movement and when the gastrocnemius lateralis and medialis EMG activity was low (under 15%).

The third finding of this study is the increased biceps femoris muscle activity from 56 to 65% of the preactivation phase of walking which corresponds to the peak muscle activity period. Individuals with CAI exhibit biceps femoris muscle activation differences compared to healthy participants prior to the initial contact on the floor. This could represent a modification of the feed-forward motor control strategy which as been suggested to be impaired for individuals with CAI (Simpson et al., 2019).

It is unclear why FOs did not modify the lower-limb EMG during the SIDE and WEDGE tasks. However, to reduce the risks of the ankle to give way or sprain, individuals with CAI exhibit a more rigid and less variable motor control pattern during locomotion (C. Herb et al., 2014; C. C. Herb & Hertel, 2015; Terada et al., 2015). It can be hypothesised that individuals with CAI exhibit a stiffer landing motor pattern, for example decreased ankle plantarflexion and knee flexion during these more challenging tasks, mitigating the FOs' EMG effects. More studies are needed to better understand the FOs' neuromuscular effects during different jump landing tasks and to infirm or confirm this hypothesis.

No significant kinematic and kinetic FOs effects were observed during any of the tasks which is consistent with previous studies that found weak or no effect during walking (Desmyttere, Hajjzadeh, Bleau, & Begon, 2018) and unilateral jump landing (Christopher, Drouin, & Houghlum, 2006) for healthy individuals.

FOs could perhaps provide their therapeutic benefits through changes in neuromuscular control rather than mechanical effects. It has been hypothesised that FOs enhance the stimulation of the plantar cutaneous mechanoreceptors providing more reliable sensory information from the foot and ankle, thus changing muscle activity during a dynamic task (Dingenen et al., 2015). However, it is still unclear to what extent neuromuscular effects without mechanical effects will translate into clinically significant patient outcomes.

The first limitation of this study is that the participants were only given a very short period of familiarisation to FOs. The biomechanical effects of FOs could possibly change with longer familiarisation and thus have greater effects on individuals with CAI. Only the immediate effects of FOs was evaluated in this study as unacceptable level of between-session biomechanical data variability was observed in a previous study (Moisan & Cantin, 2016). The results of this study should be mainly used to support or refute the need for future efficacy trials in order to investigate the long-term effects. The second limitation is that no biomechanical comparisons were made between individuals with and without CAI (copers or healthy individuals). Therefore, the biomechanical deficits associated with CAI were only assumed from the results of previously published studies. The third limitation is the EMG normalisation technique used. The readers should be cautious when comparing the results of this study with those of previously published studies that used different normalisation technique, for example with maximal voluntary isometric contractions.

5. Conclusion

FOs decrease the tibialis anterior muscle activity during the landing phase of the DROP task and increase the biceps femoris muscle activity during the preactivation phase of walking. The results of this study show that FOs' biomechanical effects are task-dependent, only affect EMG and will help to better understand their role in treatment and prevention of CAI.

Ethical approval

We confirm that any aspect of the work covered in this manuscript that has involved human patients has been approved by the appropriate ethical committees related to the institution in which it was performed and that subjects gave informed consent to the work.

Authors declaration

The author declared no conflicts of interest.

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Appendix A. Supplementary data

Supplementary data to this article can be found online at <https://doi.org/10.1016/j.ptsp.2019.08.009>.

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