Title : Kinematic, kinetic and electromyographic differences between young adults with and without chronic ankle instability during walking

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Abstract

The objective of this study was to quantify the kinematic, kinetic and electromyography differences between individuals with and without chronic ankle instability (CAI) during comfortable (CW) and fast (FW) walking. Twenty-one individuals with CAI and 21 healthy controls were recruited to walk at CW and FW speeds. The dependent variables were gluteus medius, vastus lateralis, gastrocnemius lateralis, gastrocnemius medialis, peroneus longus and tibialis anterior muscles mean activity, ankle and knee angles and moments. Kinematic, kinetic and electromyography variables were compared between groups with a one-dimensional statistical non-parametric mapping analysis. The CAI group exhibited no significant difference for ankle angles and moments compared to the control group. However, the CAI group showed less external knee rotation from 56-100% (CW) and 51-98% (FW) and more knee abduction moment from 1-6% and 7-9% (CW) and 1-2% (FW) of the stance phase. Less gluteus medius muscle activity was also observed from 6 to 9% and 99 to 100% (CW) of the stance phase for the CAI group. These results suggest proximal biomechanical compensations and will help better understand the underlying deficits associated with CAI. They also indicate that regardless of walking speeds, individuals with CAI exhibit similar differences compared to healthy participants.

1.Introduction

Lateral ankle sprains are highly common during sport-related activities (Doherty et al. , 2014). It has been estimated that 628 000 ankle sprains are seen each year in the United States emergency rooms (Waterman et al. , 2010). However, this number could represent an underestimation of the real incidence of ankle sprains, as up to 64% of people that sustain an ankle sprain do not seek professional health care (Hubbard-Turner, 2019). Among individuals sustaining an ankle sprain, 40-73% will report to incur recurrent episodes (Waterman, Owens, 2010) and seven years later, up to 72% will report residual disability (Konradsen et al. , 2002), such as chronic ankle instability (CAI) (van Rijn et al. , 2008). Individuals with CAI present deficits in strength (Hiller et al. , 2011), proprioception (Munn et al. , 2010), balance (Munn, Sullivan, 2010), postural control (Munn, Sullivan, 2010), neuromuscular recruitment (Hoch and McKeon, 2014) and gait biomechanics (Moisan et al. , 2017) compared to healthy individuals. During walking, peroneus longus muscle activity seems to be increased before (Koldenhoven et al. , 2016) and after initial foot contact for individuals with CAI compared to healthy individuals (Delahunt et al. , 2006). This muscle can also exhibit an earlier onset and a longer activity duration (Feger et al. , 2015). Even though the results are inconsistent, individuals with CAI can also present altered gluteus medius, tibialis anterior and rectus femoris muscles activity (Moisan, Descarreaux, 2017). For the kinematic parameters, previous studies reported increased rearfoot (Drewes et al. , 2009) and ankle (Delahunt, Monaghan, 2006) inversion, increased external tibial rotation (Drewes, McKeon, 2009) and decreased ankle joint dorsiflexion (Chinn et al. , 2013). Regarding kinetic parameters, individuals with CAI present a laterally deviated center of pressure and increased lateral forces under the foot compared to healthy individuals (Koldenhoven, Feger, 2016). The increased rearfoot and ankle inversion could place more load on the lateral part of the foot, and may explain the kinetic differences (Moisan, Descarreaux, 2017). The increased peroneus longus activity could represent a protective mechanism to counteract these kinematic and kinetic deficits (Delahunt, Monaghan, 2006). However, only a few studies simultaneously quantified the kinematic, kinetic and EMG differences between individuals with and without CAI thereby limiting our understanding of the underlying deficits associated with CAI.

Lower extremity muscle activity (Murley et al. , 2014), joint moments (Browning and Kram, 2007), tibio-talar plantarflexion and hallux dorsiflexion at toe off increase (Dubbeldam et al. , 2010) at faster walking speeds. In previous studies pertaining to biomechanical effects of CAI, the walking speed was either self-selected or fixed at a comfortable pace for the participants, as shown by a recent systematic review (Moisan, Descarreaux, 2017). However, even though a higher level of gait disturbance (e.g. faster walking speed) change the variability of the walking spatiotemporal parameters for individuals with CAI compared to healthy individuals (Springer and Gottlieb, 2017), the effects of walking speed on kinematics, kinetics and EMG are still unknown for this population.

The main objective of this study was to quantify the EMG, kinematic and kinetic differences between individuals with and without CAI during walking. The secondary objective was to assess if these differences change when increasing walking speed.

2. Methods and materials

2.1. Participants

Twenty-one individuals with CAI and 21 healthy individuals were recruited to participate to this study (see Table.1). Participants of the control group were gender and age matched with those of the CAI group. Inclusion criteria for the CAI group were based on the recommendations of the International Ankle Consortium (IAC) (Gribble et al. , 2014), except for the confirmation of self-reported ankle instability with one of the three recommended validated questionnaires (Ankle Instability Instrument, Cumberland Ankle Instability Tool or Identification of Functional Ankle Instability). This recommendation could not be followed as none of these questionnaires have been translated and validated in French. Participants with CAI were included if they had at least one significant ankle sprain that occurred more than one year prior to study onset and self-reported functional deficits due to ankle symptoms that were quantified by a score of respectively <90% and <80% on the Foot and Ankle Ability Measure (FAAM) Activity of daily living (ADL) and Sport (S) subscales. The limb with the less stable ankle, subjectively decided by the participants, was evaluated when they had bilateral CAI. Participants reported at least two episodes of ankle "giving way" in the last six months and/or have a feeling of instability. The exclusion criteria for both groups were as follows: having a history of lower extremity surgery or fracture that needed a surgical realignment, history of lower extremity musculoskeletal injury within the last three months, undergoing treatment for CAI or having any condition known to adversely affect gait. Furthermore, participants of the control group never sustained an ankle sprain. Prior to their participation, all subjects gave their written informed consent according to the protocol approved by the University's ethics committee (CER-16-226-07.21). Participants were recruited among the UQTR students and through the University's outpatient podiatry clinic between October 2016 and March 2017.

2.2. Instrumentation

Surface EMG (sEMG) data were collected using differential Ag sEMG electrodes (Model DE2.1, Delsys Inc, Boston, MA, USA) applied over the gluteus medius, vastus lateralis, gastrocnemius lateralis and medialis, peroneus longus and tibialis anterior. The application of the electrodes was based on the recommendations of SENIAM (Hermens et al. , 2000). Local impedance was reduced by shaving, gently abrading with fine-grade sandpaper and wiping the skin with alcohol swabs. A reference electrode was placed over the ipsilateral anterior superior iliac spine. EMG signals were differentially amplified (AMT-8, CMRR of 92 dB at 60 Hz, input impedance of 10 GW; 12-bit A/D converter) and sampled at 1000 Hz. Kinematic data were recorded at a sampling rate of 100 Hz using a three-dimensional active motion analysis system (Optotrak Certus, Northern Digital, Waterloo, Ontario, Canada). Light-emitting diodes markers were positioned on the tested limb of each subject on the following anatomical landmarks: a) greater trochanter, b) distal 1/3 of the lateral part of the thigh (3-marker rigid plate), c) lateral femoral epicondyle, d) distal 1/3 of the lateral part of the leg (3-marker rigid plate) e) lateral malleolus, f) fifth metatarsal head. A digitizing pointer was used to create virtual markers on the medial femoral epicondyle and the medial malleolus. Ground reaction forces data were recorded at a sampling rate of 1000 Hz with a force platform (Bertec Corp, OH, USA) embedded in the floor on the participants' path. Walking speed was recorded with electronic photocells timing gates (Brower Timing System, USA) positioned 1.35 meters before and after the force platform.

2.3. Protocol

First, the participants had to fill the FAAM-ADL, FAAM-S (Borloz et al. , 2011), International Physical Activity Questionnaire (IPAQ) (Criniere et al. , 2011) and the written informed consent form. They also reported the number of sustained ankle sprains and the time since the last sprain. To quantify the participants' foot morphology, the Foot Posture Index (FPI) (Redmond et al. , 2006) was used. The experimental protocol consisted of walking on a 5-meter walkway at self-selected comfortable (CW) and fast (FW) walking speeds. During the FW trials, the participants had to walk as fast as they could without running. All participants wore the same shoe model (Athletic Works, Model: Rupert), but in their proper shoe size. Walking speed order was randomized across participants. Before the dynamic trials were completed, a static trial was recorded in order to create the hip/knee/ankle/foot segments and calculate knee and ankle angles and moments. To familiarize themselves with the experimental protocol, all participants were instructed to perform 10 familiarization trials using a midgait protocol. Then, five trials were performed during which walking speed was recorded and averaged. Finally, five trials were performed. This protocol was completed at CW and FW. Trials were rejected and immediately retaken when speed exceeded ±5% of the mean speed previously determined, if the foot was not entirely on the force platform, or if participants adapted their stride length or frequency in an attempt to hit the force platform.

2.4. Data processing

Kinematic and kinetic data were processed using Visual3D software (C-motion, Inc., Germantown, MD, USA). Kinematic data were low-pass filtered using a dual-pass, fourth-order Butterworth filter with a cut-off frequency of 6 Hz. To establish an anatomical model of the lower extremity, the Calibrated Anatomical System Technique was adopted (Cappozzo et al. , 1995). The knee and ankle joints centers were respectively defined by calculating the mid-point between the medial and lateral femoral epicondyles and the mid-point between the medial and lateral malleoli. Knee joint angles were calculated using a Cardan sequence with order of X (extension/flexion), Y (adduction/abduction), and Z (internal/external rotation). As one marker was present on the foot, only the sagittal plane ankle angle (X) was calculated and the static trial ankle angle was determined as the 0° of the joint. Internal joint moments at the knee and ankle were calculated using inverse dynamics. Joint angles and moments were resolved in the proximal segment coordinate system. The ground reaction forces (GRF) data were low-pass filtered by a dual-pass, fourth-order Butterworth filter with a cut-off frequency of 50 Hz and the vertical component of the GRF, with a threshold set at 10 Newtons, was used to determine the initial and final foot contact. The EMG data were analyzed using a custom MATLAB file (Mathworks, Inc., Natick, MA). They were digitally filtered with a zero-phase lag, bi-directional, 10 to 450 Hz bandpass fourth-order Butterworth filter. Analyses were performed on the Root Mean Square (RMS) of these data, calculated with a moving window of 100 ms width with an overlap of 50 ms. RMS data of each muscle were normalized with the mean peak RMS amplitude of all FW trials.

2.5. Analysis

The Shapiro-Wilk test value was used to evaluate the baseline characteristic distribution. Mann-Whitney tests were performed with a level of statistical significance set at $p < 0.05$ to compare the CAI and control groups, as the data were not normally distributed. Walking speeds were compared between groups using one-way repeated measures ANOVA (2 groups X 2 speeds) on the log-transformed data. To evaluate the distribution of the EMG, kinematic and kinetic data, the D'Agostino-Pearson test was used. To compare the between groups effects, a curve analysis was performed using one-dimensional statistical non-parametric mapping. Each individual stance phase was normalized to 100%. The non-parametric permutation method test (SnPM) was used to compare the differences between each normalized point of the curves (Nichols and Holmes, 2002, Pataky et al. , 2015) with a threshold of α =5%. The individual probability that each supra-threshold cluster could have resulted from an equivalently smooth random process was determined. When supra-threshold clusters were observed, the highest Cohen's d effect size was calculated. The analyses were conducted using the open-source code (www.spm1d.org) with Python software (Version 2.7).

3. Results

3.1 Descriptive data

No between group difference was found for age ($p=0.76$), height ($p=0.48$), weight ($p=0.35$), body mass index ($p=0.21$) and FPI scores ($p=0.39$). The CAI group exhibited a higher IPAQ score (CAI: 2125± 1468 met-min/week vs Control: 1566 ± 1765 met-min/week, $p=0.04$) and number of sustained ankle sprains (CAI: 5.6 ± 5.4 sprains vs Control: 0 ± 0 sprain, $p<0.01$) and a decrease of FAAM-ADL (CAI: 86.4 ± 4.5 vs Control: 100 ± 0, $p<0.01$) and FAAM-S (CAI: 69.6 ± 8.0 vs Control: 100 ± 0, $p<0.01$) scores compared to the control group. There was no significant group X speed interaction ($p=0.251$). However, speed and group effects were observed according to which FW trials were faster than CW trials ($p<0.01$) and the CAI participants walked slower than the controls ($p<0.01$).

3.2 Kinematic and kinetic data

Technical difficulties with the kinematic measurements for one participant of the CAI group led to this dataset being removed from the kinematic and kinetic analyses. Graphical representations of kinematic and kinetic patterns are presented in Fig.1 and Fig.2.

During CW, no significant difference was found for the ankle sagittal angle, ankle moments, sagittal and frontal knee angles and sagittal and transverse knee moments. For the knee transverse angle, the CAI group exhibited a decreased external rotation during 56-100% of the stance phase ($p<0.01$, $d=-0.9$) compared to the control group (see Fig.1g). For the knee frontal moment, an increased abduction moment during 1-6% ($p<0.01$, $d=1.2$) and 7-9% ($p=0.02$, $d=1.0$) of the stance phase for the CAI group (see Fig.1i).

During FW, no significant difference was found for the ankle sagittal angle, ankle moments, knee sagittal and frontal angles, knee sagittal and transverse moments. For the knee transverse angle, the CAI group exhibited a decreased external rotation during 51-98% of the stance phase ($p<0.01$, $d=-0.8$) compared to the control group (see Fig.2g). For the knee frontal moment, an increase abduction moment was found for the CAI group during 1-2% of the stance phase ($p=0.02$, $d=1.0$) (see Fig.2i).

3.3 EMG data

Graphical representations of EMG patterns are presented in Fig.3. During CW, the CAI group exhibited a decreased gluteus medius muscle activity from 6 to 9% ($p=0.02$, $d=-1.0$) and 99 to 100% ($p=0.02$, $d=-0.9$) of the stance phase compared to the control group (see Fig.3.a). No significant difference was found for all other muscles at CW and all muscles at FW.

4. Discussion

This study aimed to quantify the biomechanical differences between individuals with and without CAI during shod walking at comfortable and fast speeds. No significant difference was found for the sagittal ankle angle and ankle moments when comparing the CAI and the control groups. These results contradict those of Chinn et al. (2013) that showed less ankle dorsiflexion from 42 to 51% of the gait cycle and those of Monaghan et al. (2006) that found that participants with CAI exhibited an evertor moment during the stance phase whereas the healthy participants were experiencing an invertor moment. However the contradiction between our results and those of previous studies could possibly be explained by the fact that participants in the Chinn et al. (2013) study had to walk on a treadmill and those of the Monaghan et al. (2006) study had to walk barefoot. Indeed, previous studies showed significant kinematic differences between shod/barefoot (Franklin et al. , 2015) and overground/treadmill walking (Lee and Hidler, 2008).

For the knee, a significant decrease in external rotation was observed during CW (56-100%) and FW (51-98%) for the CAI group compared to the control group. These results suggest that the knee is less externally rotated during the latter portion of the midstance until the end of the propulsion phases. These results contradict those of Drewes et al. (2009) that found increased external rotation of the tibia during barefoot treadmill walking and those of Monaghan et al. (2006) that found no knee angles difference between CAI and control participants. However, the decreased knee external rotation could be of clinical significance as there was as high as 9 degrees of difference between groups with high effect sizes. Such a difference could alter the foot position during walking and may be one of the contributing factors of the recurrent sprains for individuals with CAI.

Contrary to our finding, Koldenhoven et al. (Koldenhoven, Feger, 2016) observed an increased gluteus medius activity and hypothesized that it could represent an attempt for individuals with CAI to generate a wider base of support or stabilize the lower limb during walking. The results of the current study are inconsistent with such hypothesis and therefore further studies are needed to investigate the relationship between gluteus medius muscle function, lower limb kinematics and foot placement during walking. Furthermore, the CAI group had increased knee abduction moment during the first portion of the contact phase at CW and FW. A previous study quantified the knee moments for CAI compared to healthy participants during walking and found no difference during barefoot overground walking (Monaghan, Delahunt, 2006). However, the results of the current study could be of clinical significance as the maximum mean difference reached -124% (at 4% of the stance phase) at comfortable speed and -316% (at

2% of the stance phase). The decreased gluteus medius muscle activity and knee external rotation and increased knee abduction moment could represent proximal compensations for individuals with CAI which is consistent with a previous study that found that individuals with CAI exhibit proximal joint compensations during dynamic tasks (Terada et al. , 2014). However, it is still unknown if these biomechanical changes are a consequence or a cause of CAI. Finally, the results of this study are of importance as they improve our understanding of the underlying biomechanical deficits associated with CAI during walking.

A recent systematic review observed high heterogeneity between experimental protocols of previous studies pertaining to the biomechanical deficits associated with CAI during walking and running (Moisan, Descarreaux, 2017). Participants either walked or ran at a comfortable self-selected or predetermined speed but no study quantified the biomechanical differences when changing speed. This systematic review found that the biomechanical deficits associated with CAI were highly similar when comparing two locomotion tasks biomechanics: walking and running. The results of the current study are in line with those of the systematic review. Indeed, when comparing the kinematic and kinetic effects during CW and FW, the results were highly similar. The only discrepancy was for the EMG effects. However, even though the decreased gluteus medius muscle activity was not statistically significant during FW, a 15% decreased activity with a Cohen's d of -0.6 was observed. This difference would most likely be significant if the number of participants was greater. These suggest that regardless of the walking speed, individuals with CAI exhibit similar deficits. This could be of great impact in clinical contexts as clinicians will be able to better target the biomechanical deficits associated with CAI during rehabilitation.

The first limitation of this study is the foot kinematic model used. As one marker was positioned on the foot, only the sagittal ankle motion was quantified. It is possible that significant differences in the frontal and transverse planes were present but cannot be observed with this experimental set-up. The second limitation is that a higher IPAQ score was observed for the CAI group; it is possible that more active individuals could better compensate the biomechanical deficits associated with CAI during walking. The third limitation is that the contralateral limb's biomechanics was not assessed in this study. Individuals with CAI could exhibit biomechanical compensatory strategies to the uninjured limb but cannot be observed.

5. Conclusion

Individuals with CAI did not show any significant kinematic and kinetic differences at the ankle joint. However, an increased knee abduction moment and a decreased knee external rotation and gluteus medius muscle activity were observed for participants with CAI compared to healthy participants. These results suggest that individuals with CAI exhibit proximal compensations during walking at CW and FW and could help researchers and clinicians to better target the deficits associated with CAI during rehabilitation.

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Conflict of interest statement

The authors declare that they have no conflict of interest relating to the material presented in this article.

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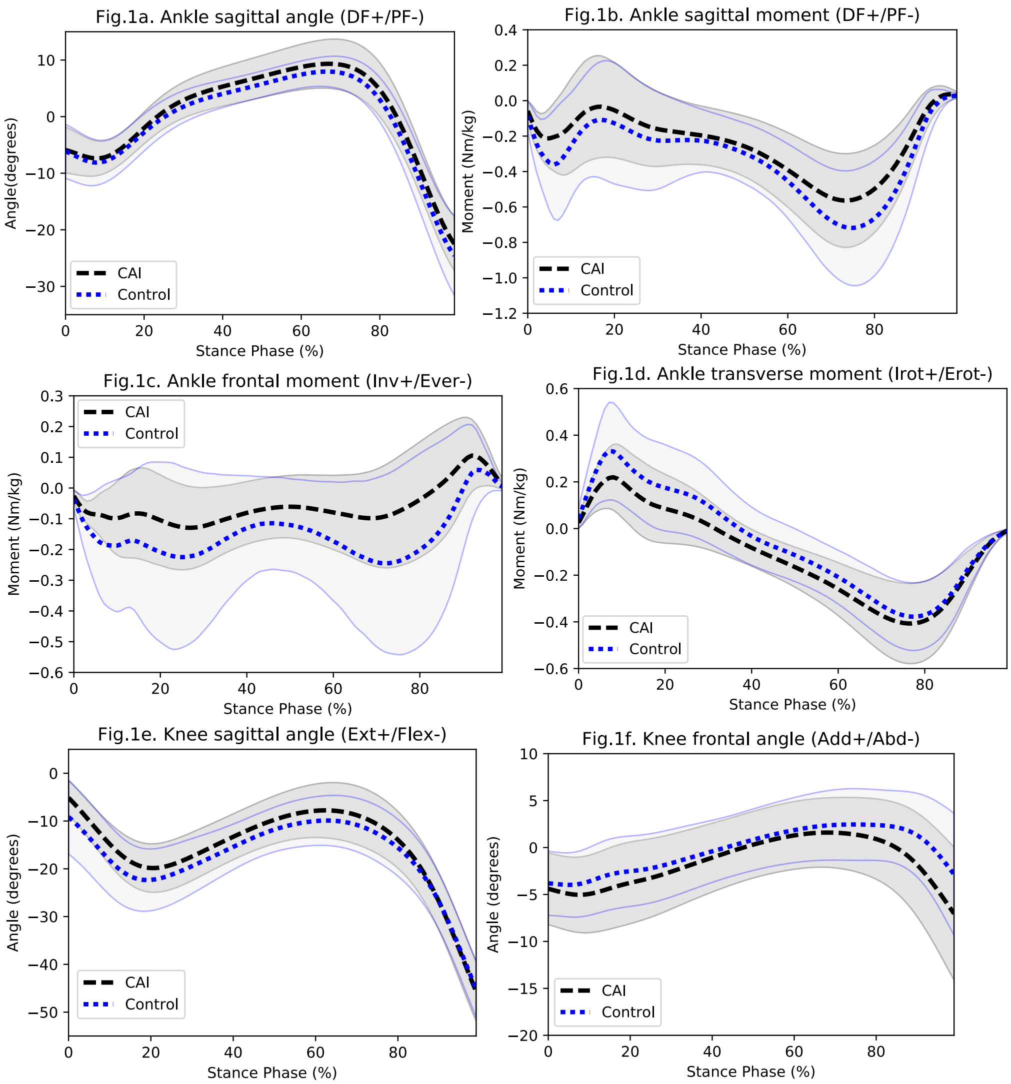
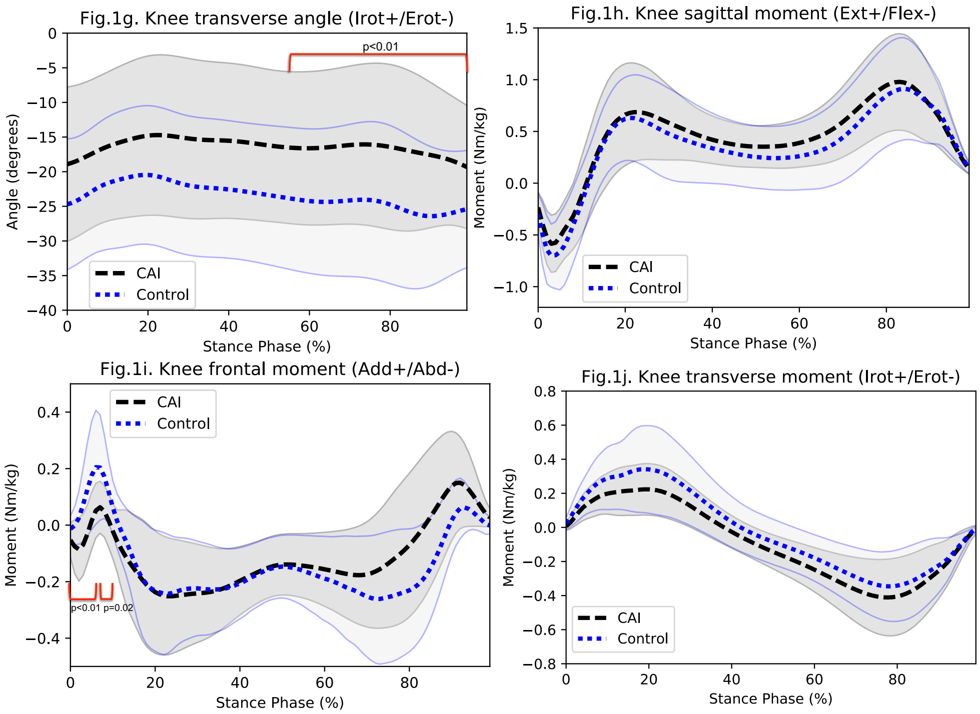
Fig.1 Ankle and knee angles and moments during comfortable walking





Captions:

DF=Dorsiflexion

PF= Plantarflexion

Inv= Inversion

Ever= Eversion

Irot= Internal rotation

Erot= External rotation

Ext= Extension

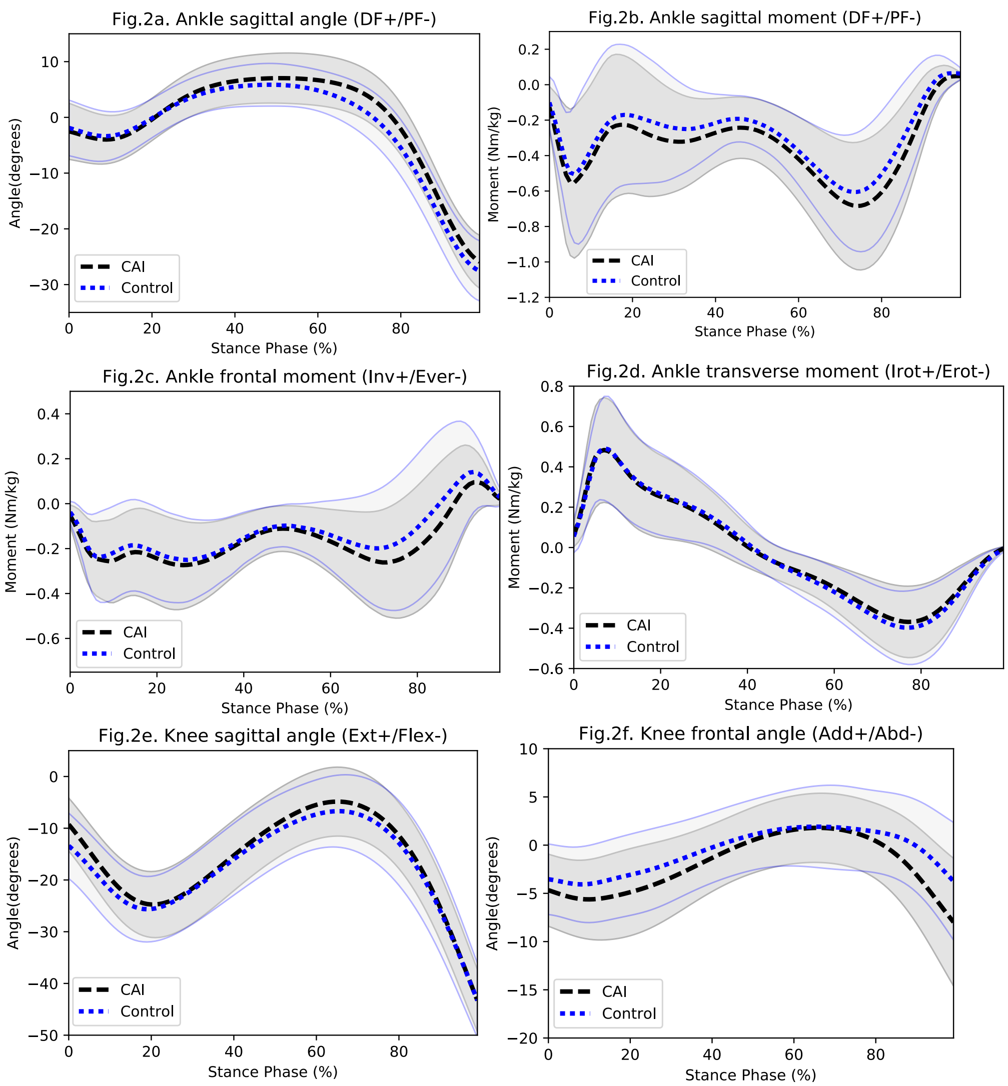
Flex= Flexion

Add= Adduction

Abd= Abduction

* Means of the CAI (black) and Control (blue) groups are respectively represented by dotted lines and standard deviations are observed between the full lines.

Fig.2 Ankle and knee angles and moments during fast walking



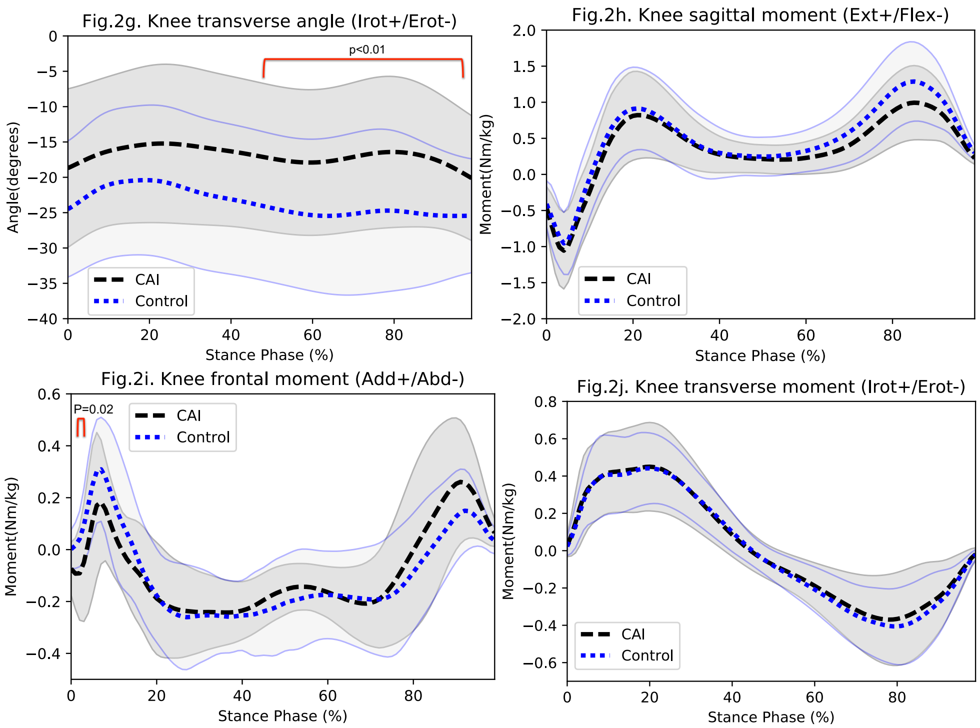


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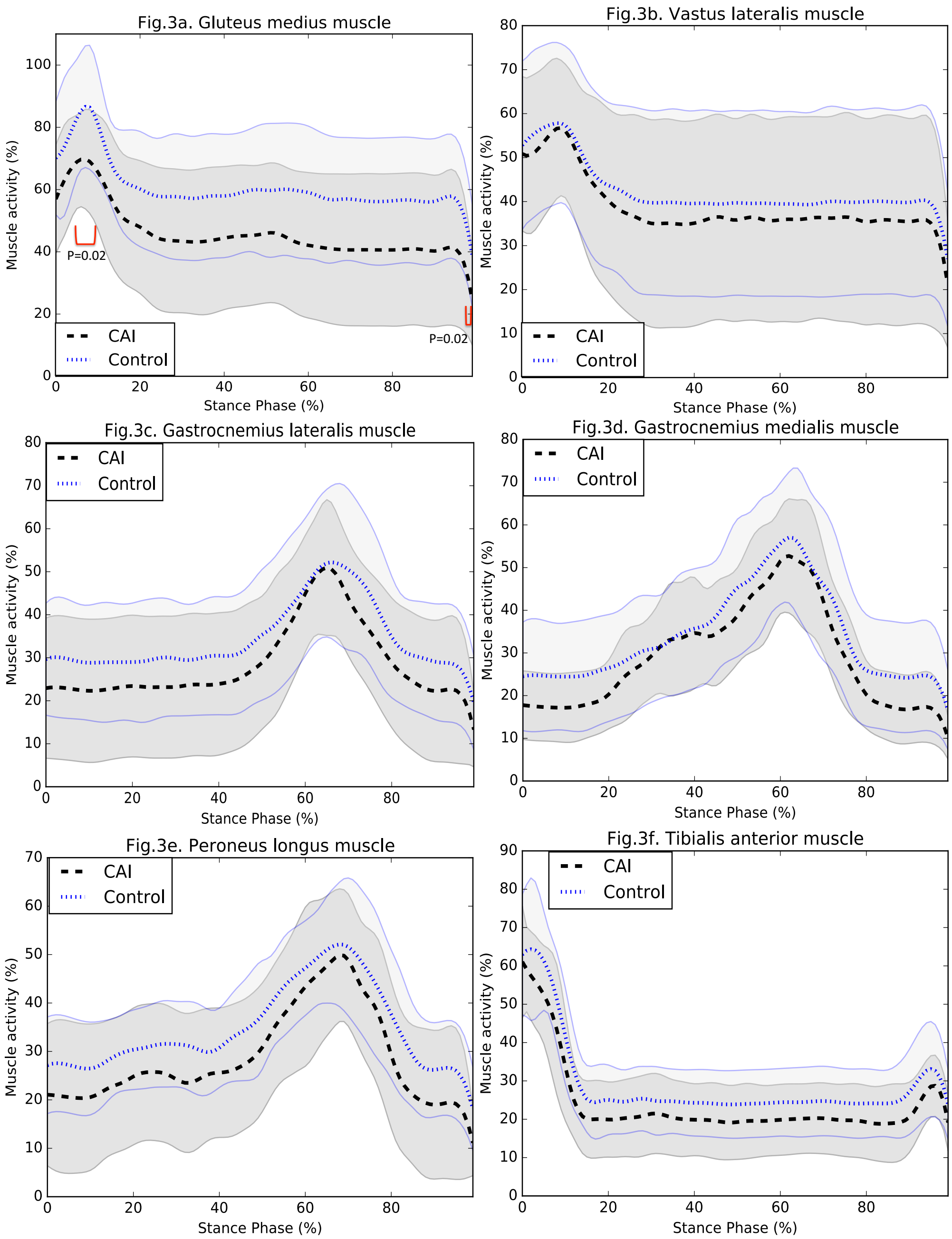
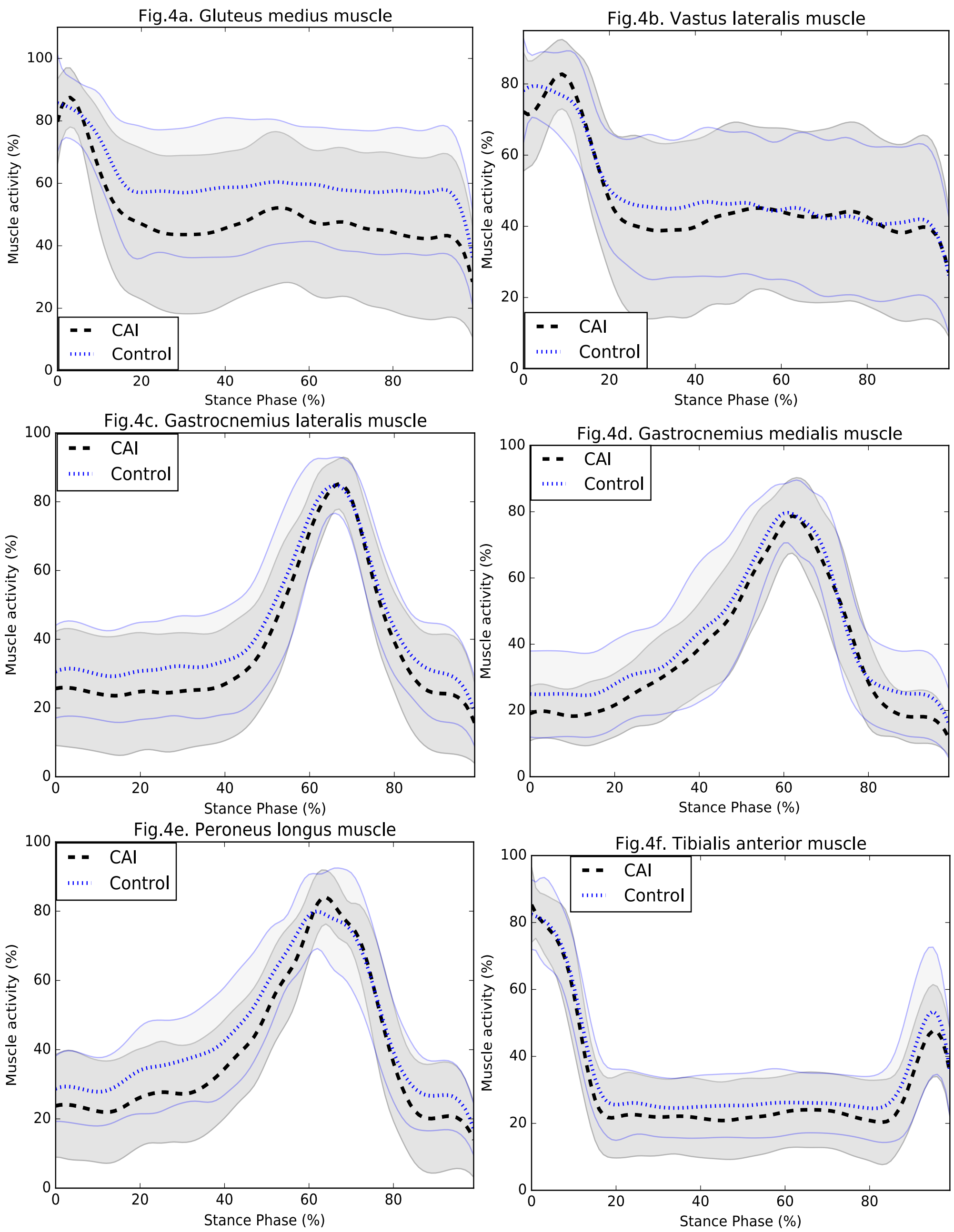
Fig.3 EMG differences between CAI and control groups during comfortable walking



* Means of the CAI (black) and Control (blue) groups are respectively represented by dotted lines and standard deviations are observed between the full lines.

Fig.4 EMG differences between CAI and control groups during fast walking



* Means of the CAI (black) and Control (blue) groups are respectively represented by dotted lines and standard deviations are observed between the full lines.

Table 1. Descriptive data

Group	CAI	Control
Gender ratio (M/F)	4/17	4/17
Age (years)	26.3 (±8.5)	25.1 (±5.3)
Weight (kg)	64.9 (±12.7)	61.7 (±12.7)
Height (m)	1.65 (±0.08)	1.67 (±0.09)
Foot posture index	3.3 (±3.8)	2.6 (±3.9)
Last sprain (yr)	2.4 (±1.9)	NA
Previous sprains	5.6 (±5.4)	0 (±0)
FAAM-ADL (%)	86.4 (±4.5)	100 (±0)
FAAM-Sport (%)	69.6 (±8)	100 (±0)
IPAQ (MET-min/week)	2125 (±1468)	1566 (±1765)
Comfortable speed (m/s)	1.38 (±0.19)	1.49 (±0.21)
Fast speed (m/s)	2.00 (±0.23)	2.12 (±0.21)

Authors' biography

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