Abstract

Background: Chronic ankle instability (CAI) has previously been linked to altered lower limb kinematics and muscle activation characteristics during walking, though little research has been performed analysing the full time-series across the stance and swing phases of gait.

Research Question: The aim of this study was to compare trunk and lower limb kinematics and muscle activity between those with chronic ankle instability and healthy controls.

Methods: Kinematics and muscle activity were measured in 18 (14 males, 4 females) healthy controls (age 22.4 ± 3.6 years, height 177.8 ± 7.6 cm, mass 70.4 ± 11.9 kg, UK shoe size 8.4 ± 1.6), and 18 (13 males, 5 females) participants with chronic ankle instability (age 22.0 ± 2.7 years, height 176.8 ± 7.9 cm, mass 74.1 ± 9.6 kg, UK shoe size 8.1 ± 1.9) during barefoot walking trials, using a combined Helen Hayes and Oxford foot model. Surface electromyography (sEMG) was recorded for the tibialis anterior and gluteus medius. Full curve statistical parametric mapping was performed using independent and paired-samples T-tests.

Results: No significant differences were observed in kinematic or sEMG variables between or within groups for the duration of the swing phase of gait. A significantly increased forefoot-tibia inversion was seen in the CAI affected limb when compared to the CAI unaffected limb at 4-16% stance ($p = 0.039$). No other significant differences were observed.

Significance: There appears to be no differences in muscle activation and movement between CAI and healthy control groups. However, participants with CAI exhibited increased inversion patterns during the stance phase of gait in their affected limb

compared to their unaffected limb. This may predispose those with CAI to episodes of giving way and further ankle sprains.

Key Words: 3D kinematics, Electromyography, Chronic ankle instability, Gait

Introduction

Lateral ankle sprains are one of the most common musculoskeletal injuries in both general and sporting populations [1]. Following an acute ankle sprain, it is suggested that 32-74% of individuals have residual symptoms such as recurrent sprains, episodes of giving way and/or perceived instability [2]. Chronic ankle instability (CAI) is defined as 'a history of recurrent ankle sprains and the sensation of giving way' [3]. Long term, links have been established between the development of osteoarthritis and a history of CAI, suggesting abnormal kinematic movement patterns adopted may increase repetitive cartilage damage to the medial ankle [4]. Greater understanding of the biomechanics associated with CAI may aid the development of preventative measures.

Walking is of high importance in daily life, and is often problematic for people with CAI who complain of giving way sensations on uneven and level surfaces [5]. Research suggests that the position of the affected ankle joint at specific time points during the gait cycle may predispose an ankle to injury [6]. This may be associated with or caused by ankle joint instability. Research analysing frontal plane ankle kinematics during walking observed increased ankle inversion that corresponded to greater ankle inversion during more sport-specific tasks such as jump-landing [7]. Gait analysis is often used in the development of rehabilitation and injury prevention protocols, therefore any changes in full body gait kinematics need to be investigated, and where possible accounted for, as these may impact not only walking but other more dynamic movements.

Previous literature investigating sEMG found hip abductor weakness to be associated with acute ankle sprains, though it is unclear whether this is a cause or an effect of the sprain [8]. Koldenhoven et al. [9] reported increased gluteus medius activation in the late

stance and early swing phase of walking in CAI participants, suggesting this may be a coping mechanism used to generate a wider base of support, or to increase lower limb stability. Decreased tibialis anterior activation was also observed resulting in increased ankle plantarflexion prior to heel strike. This loose-packed position (ligaments and the joint capsule lax and minimal joint surface contact) has been found to be unstable [10], suggesting an increased risk of ankle sprains.

Previous literature investigating CAI during walking has modelled the foot as one rigid segment [11, 12], however the foot is composed of 26 bones and 20 articulated joints with a number of complex interactions [13]. Rigid segment modelling excludes motion between different segments of the foot providing inadequate information on the biomechanics of the foot [11]. De Ridder et al. [14] appears to be the first study to analyse walking using a multi-segmental foot model, comparing the use of the Ghent Foot Model to a rigid foot model in participants with CAI, copers (no symptoms of instability after a recent ankle sprain) and control participants. Results lead the authors to conclude that the multi-segmental foot model provided greater details of the intricacies of the foot, showing differences between segments when comparing groups.

Upper body kinematic analysis should be considered when investigating changes in the lower extremities as there may be a significant relationship with changes observed in proximal segments [15]. The body is a multi-linked system with the rectus femoris, hamstrings and gastrocnemius muscles crossing the hip, knee and ankles. The kinetic chain concept suggests that movement of the trunk during landing (which accounts for 35.5% body mass) will also have an impact on motion of the hip and therefore knee and ankle [16]. To the authors' knowledge, no research has combined trunk kinematics with

a full lower limb and multi-segmental foot model to address, in combination, the possible proximal and distal differences between groups.

Prior research reports joint angles and muscle activation characteristics at discrete time points during walking [9, 12], rather than whole kinematic time-series curves. Biomechanical data is one dimensional (1D) (time and kinematic or force trajectories) therefore this may result in focus bias or missing potential significance or trends during other phases of the gait cycle [17]. Statistical parametric mapping (SPM) is a concept introduced to biomechanics from brain research [18] which enables curve analysis across the whole movement [17]. Comparison between SPM and time series analysis using confidence intervals concluded SPM to be the most suitable method for analysis of 1D data, due to increased generalisability of probabilistic conclusions (with the use of hypothesis testing techniques) and the ability to present results in a more consistent manner aiding interpretation of findings [19]. De Ridder et al. [14] used SPM to compare foot kinematics between participants with CAI, copers and controls, identifying exact time periods of significantly increased forefoot inversion within the stance phase of walking.

It is suggested that combined analysis of the trunk, hip, knee and multi-segmental foot kinematics and sEMG activation patterns across the stance and swing phases of gait will provide greater insight into possible differences that exist, not just within the foot, but across the full kinetic chain. This may provide greater insight to clinicians rehabilitating those with ankle instability and may highlight areas of importance in the reduction of future ankle sprains. The aim of this study was to compare trunk, hip, knee and multi-segmental foot kinematics and muscle activation during the stance and swing phase of walking between participants with CAI and healthy controls.

Methods

Participants

Eighteen (14 males, 4 females) healthy controls (age 22.4 ± 3.6 years; height 177.8 ± 7.6 cm; mass 70.4 ± 11.9 kg; UK shoe size 8.4 ± 1.6), and 18 (13 males, 5 females) participants with CAI (age 22.0 ± 2.7 years; height 176.8 ± 7.9 cm; mass 74.1 ± 9.6 kg; UK shoe size 8.1 ± 1.9) participated in this study. Ethical approval was granted by the institutional ethics committee prior to testing. Written informed consent was obtained from participants and a health screen questionnaire completed prior to participation. Inclusion and exclusion criteria for participation detailed in table 1., in accordance with selection criteria outlined by the International Ankle Consortium (IAC) [2].

TABLE 1 AROUND HERE

Participants were allocated into the control group or the CAI group based on results of the Identification of Functional Ankle Instability (IdFAI) questionnaire, where a score of ≥11 indicated ankle instability in accordance with IAC guidelines [2]. In the instance of bilateral ankle sprains, the involved limb was selected based on the participant's perception of greater instability. As the researcher was blinded to the questionnaire outcome, the affected limb could not be identified exclusively as either the dominant or non-dominant limb. Therefore, the affected limb was randomly matched to a control limb to adjust for the dominance effect. Limb dominance was determined by asking which leg they would use to kick a ball [10]. Mean IdFAI score for the control group was 3.71 ± 3.13 and 19.1 ± 6.25 in the CAI group's affected limb.

Protocol

Participants completed a 5-minute warm up on a cycle ergometer (Monark Ergomedic 874E, Sweden) at 60 Watts. Electromyographic data were recorded bilaterally for the gluteus medius and tibialis anterior using a DataLINK data acquisition system (Biometrics Bluetooth unit W4X8, Biometrics Ltd, Gwent, UK) sampling at 1000Hz with pre-amplified SX230-1000 electrodes. Participants' skin was prepared for electrode placement and electrodes placed in accordance with SENIAM guidelines [20]. Tibialis anterior electrodes were placed at a third of the line between the tip of fibula and the tip of medial malleolus. Gluteus medius electrodes were placed half way between the crista iliaca and the trochanter. For each muscle, three maximal contractions were performed for a 5 second duration, 1-minute rest between trials. Peak activation of the three trials was identified as the maximum voluntary isometric contraction (MVIC) which was used to allow comparison between participants' sEMG data and to voluntary contractions to inspect for crosstalk. Gluteus medius MVIC was performed in side lying with the participant maximally abducting their hip (positioned mid-range) into a rigid strap positioned just above the knee [21]. Tibialis anterior MVIC was performed in a seated position and the participant maximally dorsiflexing and inverting their foot against a rigid strap [21].

Motion analysis data were recorded using an Owl Digital Real Time 10 camera system (Motion Analysis, Santa Rosa, California) sampling at 200 Hz. The motion analysis system was calibrated as per the system instructions. Passive reflective markers were attached to the participant using double-sided tape, in accordance with the Helen Hayes marker set [22] combined with the Oxford foot model [11, 23]. Marker and electrode placement were performed by the same person for all participants.

Participants were instructed to walk at their normal walking speed through the calibrated capture volume. Pace was not controlled, as this was deemed to be unnatural and has been previously shown to impact on stride time variability due to increased central nervous system involvement [24]. Participants walked barefoot 3.5 m before data were collected [25] and proceeded for 7 m across the walkway. Walking speed was recorded using pelvis segment velocity. Barefoot walking was used in accordance with the method of De Ridder et al. [14] and due to the number of markers on the foot. Participants performed a familiarisation until they were comfortable with the movement, before recording three trials for analysis [26]. Trials were deemed successful when all tracking markers were in view of the cameras and where there was no evidence of gait modification. Trials where gait modification occurred were discarded and re-tested.

Data and Statistical Analysis

Data were inspected using Cortex software (Cortex-64 5.3.1.1543, Motion Analysis Corporation, Santa Rosa, California) before importing into Visual 3D (Visual3D v6 x64, C-motion, Germantown, Maryland). Data were smoothed using a 6 Hz Butterworth filter. Initial contact was determined using the method proposed by O'Connor et al. [27], which creates a new signal by calculating the midpoint between the posterior inferior heel marker and the toe marker (between 2nd and 3rd metatarsal heads). The first derivative was calculated on the vertical component of the signal. Event markers were created at the minimum value for heel strike and maximum value for toe off. Electromyographic data were root mean squared by a moving window of 100 ms and normalised to MVIC. Visual inspection of the data identified noise in the signal for two of the participants, warranting their sEMG data be removed. To maintain pre-experimental research design, matched controls assigned to the two participants also had their sEMG data removed.

Kinematic and sEMG data were exported for the stance (heel strike to toe off) and swing (toe off to heel strike) phases into MATLAB R2015a (The Math Works, Natick, Massachusetts) to perform SPM analysis.

Kinematic data were exported for forefoot-hindfoot angle (FFHFA), forefoot-tibia angle (FFTBA), hindfoot-tibia angle (HFTBA), hip, knee and trunk angles in the sagittal, frontal and transverse planes of motion. So not to eliminate inherent variations in foot morphology, data were not normalised against a reference segment [14, 23]. Data were analysed using SPM in MATLAB (SPM1D open-source package, spm1d.org). Normality was tested using a D'Agostino-Pearson's test. A matched control limb was compared to the CAI groups' affected limb using an independent-samples T-test ($\alpha= 0.05$). The unaffected and affected limb of the CAI group were compared using a paired-samples T-test ($\alpha= 0.05$). A matched control limb was compared to CAI groups' unaffected limb using an independent-samples T-test ($\alpha= 0.05$).

Results

Independent-samples T-tests revealed no significant differences ($p > 0.05$) between groups for age, stature, mass, or shoe size. An independent-samples T-test reported no significant difference in walking velocity when comparing the control group (1.20 ± 0.15 m.s^{-1}), and CAI group (1.18 ± 0.09 m.s^{-1}).

No significant differences were observed in FFHFA, FFTBA, HFTBA, hip, knee, or trunk angles in the sagittal, frontal, or transverse planes of motion, in the stance or swing phase, between the matched control and the CAI groups affected limb. No significant differences were observed in the gluteus medius or tibialis anterior muscle activation in either phase of gait between the matched control and the CAI groups affected limb.

A significant difference was reported between the CAI groups' unaffected and affected limb in the FFTBA in the frontal plane, where increased inversion was observed in the affected limb at 4-16% of the stance phase (mean difference = 3.07°, peak difference = 3.24°, p = 0.039, Figure 1.). No other significant differences were reported for FFTBA. Furthermore, no significant differences were noted between FFHFA, HFTBA, hip, knee, or trunk angles or in muscle activation of the tibialis anterior and gluteus medius between the unaffected and affected limbs at any time point. Finally, no significant differences were observed between the CAI groups' unaffected limb and the control groups' limb (matched for dominance) in any of the recorded variables in either the stance or swing phases of movement.

FIGURE 1 AROUND HERE

Discussion

The aims of this study were to explore the differences in kinematics and muscle activation patterns between CAI participants' unaffected and affected ankles and to compare the same variables to a matched control group throughout the gait cycle.

Increased FFTBA inversion was found in the affected limb of the CAI group when compared to its unaffected counterpart at 4-16% stance. This finding is of particular clinical interest, supporting previous hypotheses that participants with CAI may exhibit altered joint position sense and proprioceptive awareness [28]. Increased inversion at ground contact decreases bony restrictions of the foot-ankle complex, thus, when loaded with bodyweight increases inversion torque and joint susceptibility to injury [28]. The early period of the stance phase is beyond conscious control [6, 12], thus increased inversion places the ankle in a position of increased vulnerability at heel strike,

potentially predisposing the affected limb to further ankle sprains and episodes of giving way. Whilst not within the remit of this study, differences in angular displacement associated with CAI may be exacerbated during more dynamic movements e.g. cutting, single/double leg landing, running, or when walking on uneven surfaces, as research has previously shown increased kinematics in walking often correspond to increased kinematics during more dynamic sporting activities [7].

The lack of significant differences at the hip or knee, between groups, in the frontal, sagittal or transverse planes of motion in the current study is consistent with the findings of Monaghan et al. [12], who found no significant differences in hip and knee kinematics between participants with CAI and healthy control participants from 100 ms pre-heel strike to 200 ms post-heel strike. Within the current study, trunk kinematics were measured in all three planes, however, no significant differences were identified between groups suggesting that no proximal adaptations took place within the CAI group during walking.

No significant differences were observed in tibialis anterior or gluteus maximus muscle activation between groups during gait. This is contrary to the findings of Hopkins et al. [10] who when reporting discrete peak value data, observed an increase in tibialis anterior activation from 15-30% and 45-70% of stance, which they speculated was a motor strategy to maintain a more dorsiflexed, stable position in the affected limb compared to a dominance matched control limb. Methodological differences exist between the current study and the study by Hopkins et al. [10] as participants walked shod rather than barefoot as in the present study. Decreased muscle activation patterns have previously been observed in barefoot walking compared to shod walking [29]. Hopkins et al. [10] also examined tibialis anterior activation whilst walking on a treadmill

rather than over ground. These differences in methodological approaches may account for the differing results between the two studies. Koldenhoven et al. [9] recorded significantly higher gluteus medius muscle activation in the final 50% of stance and the first 25% of the swing phase, when compared to healthy participants, however, this was again performed shod on a treadmill, making comparisons with the current study difficult. Previous studies have found different muscle activation patterns and sagittal plane motion with treadmill walking compared to over ground walking [30]. Therefore, the results from this study may prove a more valid representation of the everyday task of over ground walking. Furthermore, comparison to previous research may not be entirely valid due to the different statistical analysis used [9, 10, 29, 30]. It is important to note that grouping of participants was purely through the inclusion criteria outlined in the IAC guidelines and with use of the IdFAI questionnaire [2], no other discriminative measures e.g. Beighton score for hypermobility were used. This may be a limitation although further research is required to establish this.

Our study observed no differences in biomechanics between healthy controls and participants with CAI, however we did find differences between affected and unaffected limbs of the CAI group. This may suggest greater inversion during the stance phase is a direct result of the ankle sprain or a predisposing factor for injury. Early gait reeducation could be warranted as individuals return to walking; we make this statement with caution as a prospective study is warranted to truly determine whether greater inversion is present prior to or as a result of injury.

This study analysed kinematic and electromyographic parameters to determine differences in movement patterns and muscle activations. Future research should identify the impact of CAI on kinetic parameters using full curve analysis to identify

differences between groups. Further research should use these analysis methods to examine dynamic movements such as change of direction, both single and double leg landing and running gait. Analysis of additional muscle sEMG signals may also provide greater understanding of potential differences between groups. In particular muscles such as the peroneals which may be a causative factor of the differences observed in FFTBA frontal plane kinematics.

Conclusion

Participants with CAI exhibited increased inversion patterns during the stance phase of gait in their affected limb compared to their unaffected limb. This change in movement pattern may predispose those with CAI to episodes of giving way and further ankle sprains. Increased inversion may also be a significant risk factor in more dynamic movements, thus further research should investigate these using a multi-segmental foot model. Incorporating kinetic variables into this analysis may also be beneficial to determine differences in ground reaction forces and moments.

Conflict of interest

None.

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