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Sagittal Plane Gait Kinematics in Individuals With Chronic Ankle Instability

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Single joint kinematic alterations have been identified during gait in those with chronic ankle instability (CAI). The purpose of this study was to compare sagittal plane hip, knee, and ankle kinematics during walking in participants with and without CAI. Twelve individuals with CAI and 12 healthy individuals walked on a treadmill at 1.5 m/s. Three-dimensional kinematics were analyzed using mean ensemble curves and independent *t* tests. Participants with CAI demonstrated less lower extremity flexion during the absorption phase of stance and the limb placement phase of swing, which may have implications for limb placement at initial contact.

Ankle sprains are among the most common orthopedic injuries sustained within the general population.¹ Over 600,000 ankle sprains are treated annually in United States emergency rooms alone.² Approximately one in three individuals who sustain a single acute ankle sprain develop a health condition known as chronic ankle instability (CAI), which is characterized by bouts of recurrent ankle instability resulting in multiple future ankle sprains.^{3,4} In addition to the trauma associated with acute bouts of ankle instability and sprains, CAI has been linked to decreased physical activity,⁵ development of ankle osteo-

arthritis,⁶ and functional loss,^{7,8} indicating this condition is associated with long-term negative sequelae over the lifespan.⁹

The development of CAI is thought to be a multifactorial phenomenon based on alterations in the mechanical structures surrounding the ankle complex and functional deficits in sensorimotor control that may occur throughout the lower extremity.¹⁰ The interaction between mechanical alterations in joint laxity, arthrokinematics, and degenerative structural changes with functional impairments in proprioception, neuromuscular control, and postural control are likely responsible for the repetitive ankle trauma associated with CAI.^{4,10} The negative consequences of this interaction may be most eminent during dynamic activities, such as gait, which require the coordination of several joints and segments during cyclic transitions from loaded to unloaded conditions while maintaining a base of support.^{4,11}

Several alterations in walking gait kinematics have been identified in people with CAI.^{11–13} These alterations included increased rearfoot inversion at initial contact,^{11,12}

KEY POINTS

Few studies have examined kinematics of the hip and knee during gait in individuals with chronic ankle instability.

During walking gait, participants with chronic ankle instability demonstrated substantially lower extremity flexion during the absorption phase of stance and this corresponded to an increased peak vertical ground reaction.

In the swing phase of walking, those with CAI demonstrated substantially less flexion during the limb placement phase of swing.

Examining the sagittal plane summary profile of the hip, knee, and ankle provides more robust information about the absorption capabilities of those with chronic ankle instability compared with examining any one individual joint.

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Participants

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allow participants to adjust to the treadmill before data collection. Once the target speed of 1.5 m/s was maintained for 1 min, a 30-s trial of walking was recorded.

Procedures

To assist in accurately placing the retroreflective markers for three-dimensional motion capture, all participants were barefoot, wore close-fitting shorts, and females wore tank tops and males wore no tops. A custom anatomical marker set was used to capture hip, knee, and ankle motion during gait. Retroreflective spherical markers (10 mm) were placed bilaterally over the anterior and posterior superior iliac spine, greater trochanter, lateral and medial femoral condyle, medial and lateral malleoli, and head of the first metatarsal using adhesive tape. After being outfitted with retroreflective markers, a static trial was recorded in an anatomical standing position followed by walking on the treadmill.

Data were reduced using custom written code in Matlab 7.9.0 (Mathworks Inc, Natick, MA, USA). From the 30 s of data capture, approximately 20 strides of the involved limb were recorded. Each stride was resampled to 101 frames to represent each percent of stride (0% representing initial contact, 100% instant before initial contact on the same limb). For each participant, a reference angle for each kinematic variable of interest was determined from the recorded static trial and was subtracted from the angles recorded during walking.^{11,12} Using the custom MatLab code, five nonconsecutive strides were selected and averaged for each participant. Kinematic variables were reduced from the three-dimensional angles for (1) knee flexion/extension, (2) hip flexion/extension, and (3) ankle dorsiflexion/plantar flexion using a previously established method.³² In addition, the sagittal kinematic angles from each joint were summed for each data point of the gait cycle to produce a summary lower extremity profile of flexion and extension for each of the 101 data points. Flexion angles (hip flexion, knee flexion, and ankle dorsiflexion) were defined in the positive direction whereas extension was defined in the negative direction (hip extension, knee extension, and ankle plantar flexion).

Statistical Analysis

To determine meaningful differences in kinematics, a mean ensemble curve analysis^{11,12} with standard error (*SE*) at each of the 101 data points was calculated across the entire gait cycle. *SE* was selected over other types of intervals as to not exclude areas where

subtle differences may be present, which would be washed out by larger measures of variance. Group differences were considered periods of at least three consecutive data points where the *SE* intervals for each group did not overlap. The percentages of the gait cycle associated with nonoverlapping intervals were pooled for each participant. The pooled value for each participant was used to create group means, which were then compared using independent *t* tests. The level of significance was set *a-priori* at $\alpha \leq .05$ for all analyses. Effect sizes (ES) were calculated based on the mean difference and pooled standard deviation using a bias-corrected *Hedge's g* with 95 % confidence intervals (CI).³³ ES were interpreted as weak (0–0.39), moderate (0.40–0.69), and strong (≥ 0.70).³⁴ All analyses were conducted using Excel 2007 (Microsoft, Redmond, WA, USA).

Results

Individual Joint Analysis

Based on the average vertical ground reaction forces of all participants, the stance phase was 0–63 % whereas swing phase was 64–100 % of the gait cycle. Individuals in the CAI group demonstrated decreased knee flexion from 69–98 % of the gait cycle (Healthy: $42.8 \pm 5.9^\circ$, CAI: $35.2 \pm 6.6^\circ$; $p = .007$; ES = 1.16 ± 0.87) and decreased hip flexion from 70–100 % of the gait cycle (Healthy: $16.6 \pm 3.0^\circ$, CAI: $13.4 \pm 3.6^\circ$; $p = .01$; ES = 0.93 ± 0.84). Individuals with CAI also demonstrated decreased hip flexion from 0–43 % of the gait cycle, however it was not statistically significant (Healthy: $7.7 \pm 3.4^\circ$, CAI: $5.5 \pm 4.0^\circ$; $p = .16$; ES = 0.57 ± 0.82). Finally, those with CAI demonstrated decreased ankle dorsiflexion from 15–32 % (Healthy: $3.9 \pm 2.6^\circ$, CAI: $1.7 \pm 3.5^\circ$; $p = .10$; ES = 0.66 ± 0.82) and 90–97 % (Healthy: $0.9 \pm 2.3^\circ$, CAI: $-1.1 \pm 2.9^\circ$; $p = .07$; ES = 0.75 ± 0.82) of the gait cycle; however, it was also not statistically significant. The group mean ensemble curves are presented in Figures 1, 2, and 3.

Lower Extremity Summary Kinematic Analysis

From the summary analysis, a distinct pattern of flexion and extension across the lower extremity emerged and was consistent across both groups, however there were substantial differences between groups (Figure 4). In the stance phase, individuals in the CAI group displayed significantly less flexion from initial contact to 45% of the gait cycle (Healthy: $24.1 \pm 6.8^\circ$, CAI:

Figure 1 Ankle dorsiflexion/plantar flexion kinematics during the gait cycle for healthy ($n = 12$) and chronic ankle instability ([CAI] $n = 12$) participants. The percent of gait cycle is from initial contact (0%) to the instant before the subsequent initial contact on the same limb (100%). The solid line represents the group mean while the dashed lines represent the mean $\pm$ SE.

Figure 2 Knee flexion/extension kinematics during the gait cycle for healthy ($n = 12$) and chronic ankle instability ([CAI] $n = 12$) participants. The percent of gait cycle is from initial contact (0%) to the instant before the subsequent initial contact on the same limb (100%). The solid line represents the group mean while the dashed lines represent the mean $\pm$ SE.

$17.6 \pm 6.5^\circ$; $p = .02$; $ES = 0.94 \pm 0.84$), which corresponded to the time windows for the heel and ankle rockers, representing the absorption phase of stance. There were no significant differences between groups

from 45% to 71% of the gait cycle, however there were substantial differences again from 72% of the swing phase to initial contact (Healthy: $53.5 \pm 6.4^\circ$, CAI: $44.8 \pm 5.8^\circ$; $p = .002$; $ES = 1.38 \pm 0.89$).

The altered hip and knee kinematics in those with CAI have substantial implications, but this is the first study to determine these alterations. Previous^{12,35,36} reports which have examined hip and knee gait kinematics of those with CAI did not identify significant differences, however these studies primarily examined kinematics immediately before and during stance phase. In the current study, many of the alterations in hip and knee kinematics were identified during the mid- and terminal aspects of swing, which were not examined in previous studies. Another key difference in these past studies is that they examined overground walking, while the current study used a treadmill with a standardized speed. Overall, the more extended hip and knee kinematics during swing phase and the overall summary decrease in flexion range of motion in the latter part of swing suggests a decreased ability to control motions related to limb placement in preparation for the transition from an unloaded to loaded limb in those with CAI. The more extended hip and ankle during the stance phase and the overall summary decrease in the early part of stance suggests individuals with CAI may have a decreased ability to effectively absorb forces arising from initial contact which may influence the ability to attenuate forces and maintain a stable base of support during stance.^{6,37} In a secondary analysis of the vertical ground reaction force profile of these groups, we found that in the early part of stance the CAI group demonstrated significantly higher normalized vertical peak ground reaction force compared with the healthy group (Healthy: 0.8 ± 0.1 N/kg, CAI: 0.9 ± 0.1 N/kg; $p = .03$; ES = 0.97 ± 0.85). These findings, while not part of the initial design of the study, help to substantiate the hypothesis that the summary sagittal kinematic profiles between the groups provide insight into the ability to attenuate the forces arising from ground contact. In addition, these results support previous research which identified difficulty with foot clearance during swing,¹⁵ which, when combined with the inability to control limb placement and reduce ground reaction force, may promote episodes of giving way due to tripping or stumbling in more unpredictable environments. When combined with the self-reported functional deficits of the CAI group as reported via the FAAM-ADL and the FAAM-Sport, it is quite possible that these kinematic alterations provide insight into the underlying factors related to functional deficits experienced by those with CAI.

While significant differences were detected in hip and knee kinematics, no statistically significant differ-

ences were detected in ankle dorsiflexion. Although significant differences in dorsiflexion were not identified between groups, the CAI group demonstrated strong trends toward decreased dorsiflexion during stance and swing phases compared with healthy individuals. This finding supports previous research in which those with CAI exhibited approximately 3° less dorsiflexion from 42–51 % of the walking gait cycle.¹⁴ While this study¹⁴ examined walking gait on a treadmill, it differed from the methods of the current study by using a shod condition. Although the mean differences in this study were only 2.2° , the moderate-to-strong ESs indicate these trends may have clinical implications. Specifically, these trends may reflect functional deficits in force absorption, the transition to propulsion,^{22,23} and the transition to a loaded condition for people with CAI.¹⁵ When considering the summary findings, it appears that the cumulative effect of ankle dorsiflexion combined with knee and hip flexion profiles elucidate the functional deficit in those with CAI to a greater extent than any single joint alteration alone. There is also evidence to suggest that the amount of dorsiflexion available influences knee flexion angle during landing as well as the magnitude of the vertical ground reaction force during the absorption phase.^{21–24} As well, those with CAI demonstrate reduced weight bearing dorsiflexion²⁵ as well as a talar positional fault³⁸ due to posterior capsular restrictions in the talocrural joint. Therefore, future studies that examine gait in individuals with CAI should also consider assessing alterations in dorsiflexion range of motion not only to simply increase motion at this individual joint, but to free up the entire lower extremity absorption capabilities.

The retrospective design of this study does not permit a causal link to be established between CAI and these alterations in gait. Another limitation is that all participants walked barefoot on a treadmill. While using a treadmill allowed for controlled speed and efficient capture of multiple trials, barefoot walking on a treadmill may not be directly generalized to the common injury mechanisms associated with recurrent ankle sprains. We used a set walking speed (1.5 m/s) for all participants which may not have represented the preferred walking speed for each individual and potentially created an artificial environment to some extent. While the methods used in this study enabled more precise marker placement and control over certain walking parameters, examining shod or overground gait may have elicited different results in

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