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LOWER LIMB KINEMATICS OF SUBJECTS WITH CHRONIC ACHILLES TENDON INJURY DURING RUNNING

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This study examined the kinematic differences between subjects who had a history of chronic Achilles tendon (AT) injury and matched controls during running. Eleven subjects from each group ran barefoot (BF) and shod at self-selected speeds on a treadmill. Three-dimensional angles describing rearfoot and lower limb motion were calculated throughout

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stance. Five footfalls were obtained for each subject and condition. Pairwise comparisons revealed greater eversion, ankle dorsiflexion and less leg abduction during stance in the AT group compared with controls. Running kinematics were exaggerated in shod compared with BF conditions, as expected from previous research. The differences between conditions were more exaggerated in AT subjects compared with control subjects. Further analysis using a curve-based approach is recommended.

Keywords: rearfoot, lower limb, three-dimensional, pronation

INTRODUCTION

Improved awareness of the importance of physical activity in maintaining a healthy, balanced lifestyle has led to increased participation in running and jumping activities. The repetitive nature of these movements can lead to overuse injuries in both recreational and competitive athletes. Injuries to the lower extremity are most common with the Achilles tendon (AT) affected in 5%–18% of cases (Gross, Davlin, and Evanski 1991; Kaufman et al. 1999; Taunton et al. 2002). The literature contains several comprehensive reviews of AT injury (Cook, Khan, and Purdam 2002; Kader et al. 2002; Paavola et al. 2002; Schepsis, Jones, and Haas 2002; Smart, Taunton, and Clement 1980). The mechanisms of this condition are multifactorial and depend on a wide range of intrinsic and extrinsic factors, including anthropometric factors, structural alignments, muscle and joint stiffness, concentric and eccentric strength, training errors, running surfaces, and footwear (Paavola 2001). Ground contact may result in rapid and violent motions of joints in the foot and lower leg, soft tissue stresses, and high muscle forces (Shorten 2000). It is proposed that these rapid movements may cause microtears within the tendon (McCrory et al. 1999), leading to tissue degeneration and overuse pathologies (Shorten 2000).

Two specific mechanisms have been suggested for AT injury. The first involves high levels of pronation, which has been widely linked to many musculoskeletal injuries including AT injury (Bartlett 1999; Clement, Taunton, and Smart 1984; Smart et al. 1980). Structural and biomechanical abnormalities such as rearfoot varus or an exaggerated forefoot varus may lead to compensatory pronation. When prolonged, this may result in contradictory tibial rotational forces and a wringing-like action to the tendon (Clement et al. 1984). These misalignments have been found in 50%–56% of individuals with AT injuries (Clement et al. 1984; Peterson and Renstrom 2001); however, a definite link attributing pronation as a cause of injury has not been found. This is supported by the absence of injury in some individuals who display high levels of pronation (Payne 1999), although an objective value for "excessive" levels has yet to be defined. The second proposed mechanism involves rapid alternations between concentric and eccentric contractions of the triceps surae during ground contact (McCrory et al. 1999).

Clinical research often compares control subjects with those with specific functional or structural factors such as excessive pronation (Engsberg 1996; McClay, and Manal 1998) or high and low arches (Williams, McClay, and Hamill 2001). Few studies have focused on kinematic (Harrison, Laxton, and Bowden 2001; McCrory et al. 1999) or kinetic and EMG-based analysis (Baur et al. 2004; McCrory et al. 1999) in AT subjects. There is a need for controlled studies that examine the link between pronation and the occurrence of AT injury. The aim of this study was to compare lower limb kinematics during BF and shod running in subjects with a history of chronic AT injury with uninjured controls. Only AT subjects where a high level of pronation during stance was the likely mechanism of injury were included. It was hypothesised that (i) the AT group would show greater eversion (EV) and ankle dorsiflexion (ADF) during stance compared with controls in both running conditions, and (ii) the magnitude of motion would be greater in shod running compared with BF running.

METHOD

Subjects

Twenty-two subjects provided written, informed consent to participate in this study in accordance with human research procedures. Eleven subjects with a history of chronic, low-grade AT injury were recruited from the Podiatry Department private patient files at the University of Salford (1 female, 10 males; age = 39.6 ± 7.7 years; height = 1.74 ± 0.05 m; weight = 71.9 ± 7.3 kg). In the year prior to testing, all AT subjects visited the collaborating podiatrist for a consultation. The podiatrist carried out a series of clinical observations, including qualitative analysis of BF running, calcaneal alignment in relaxed standing position and supine neutral subtalar position. All subjects displaying levels of pronation during running, which based on the podiatrist's judgement, were likely to be related to the clinical presentation of AT injury were invited to participate in the study. Those with AT injury but who displayed a rigid foot type with little visible pronation during running were excluded. After the initial consultation, all subjects were provided with custom-made orthoses, which successfully relieved the symptoms of injury. These devices were removed for the duration of the study. Eleven control subjects with no history of

AT injury were recruited from local running clubs (1 female, 10 male; age = 45.2 ± 8.1 years; height = 1.77 ± 0.05 m; weight = 77.9 ± 11.6 kg). Prior to testing, the podiatrist performed the clinical tests outlined above on both legs of the AT and control subjects to determine foot alignment. Control subjects were matched as closely as possible to the AT subjects for age, gender, height, and weight; however, they were not matched for foot type or levels of pronation. All subjects had good fitness levels, no injuries at the time of testing, and no unusual running patterns.

Experimental Set-up and Testing

All subjects completed a questionnaire providing details of their sports participation and stretching habits, while the AT subjects provided additional information about the injury. Subjects completed a familiarisation session, which involved running on a treadmill until they were fully accustomed to the mode of running (minimum 4 minutes). A marker set-up similar to that used by Clarke et al. (1983) was used. Retroreflective markers were placed on posterior and lateral aspects of both lower extremities as follows: two markers on the posterior aspect of the shoe/rearfoot bisecting the heel, two markers bisecting the posterior shank (one on the AT, one below the belly of gastrocnemius), markers on each of the fifth metatarsals, lateral malleoli, fibular heads, and greater trochanters. Three-dimensional kinematic data were captured using eight synchronous ProReflex MCU240 cameras, operating at a sampling frequency of 200 Hz and Qualysis motion capture system (Gothenburg, Sweden). Cameras were located in an arc around the posterior and lateral aspects of the treadmill. A retroreflective marker was placed on the rigid frame at the front of the treadmill; this marker indicated the position of the treadmill surface on the motion capture screen. The podiatrist placed all subjects in subtalar neutral position prior to each condition, and marker coordinates were obtained. Testing procedures required subjects to run at self-selected comfortable speeds in BF (2.5 ± 0.4 $m.s^{-1}$) followed by shod conditions (2.8 ± 0.2 $m.s^{-1}$). The BF running speeds were always the same or lower than the speeds chosen for shod running. Data capture took place during the third minute of continuous running, and subjects took full recovery before the shod condition.

Data Analysis

Raw marker coordinate data were exported from Qualysis and imported as scaled coordinates into the Peak Motus ™ analysis system (Peak Performance Technologies, Englewood, CO, USA). The three-dimensional frontal and sagittal plane angles described in Table 1 were calculated. These angles were exported to Microsoft Excel, where they were calculated

Table 1. Angles That Were Calculated to Describe Rearfoot and Lower Leg Motion

Angle	Definition
Leg abduction (ABD) angle	Angle between the lower leg and the ground on the medial side as viewed from posterior; indicates level of varus/valgus of lower leg.
Calcaneal angle	Angle between the rearfoot and the ground on the medial side as viewed from posterior; indicates inversion/eversion of rearfoot.
Eversion (EV) angle	Angle between the rearfoot and the lower leg on the medial side as viewed from posterior; indicates inversion/eversion of rearfoot relative to the lower leg.
Ankle dorsiflexion (ADF) angle	Anatomical joint angle between fibular head, ankle and 5th metatarsal; indicates level of ankle dorsiflexion/plantarflexion.
Knee flexion (KF) angle	Anatomical joint angle between greater trochanter, fibular head and ankle; indicates level of knee flexion/extension.

relative to subtalar joint neutral position taken prior to the dynamic trials. Heel strike (HS) and toe-off (TO) events for individual footfalls were determined from the displacement of the treadmill marker. Stable z-coordinates of the marker indicated the flight phase when there was no impact force on the treadmill, while a steep decrease in the z-coordinates indicated impact between the foot and the treadmill. The average of the stable z-coordinates during the flight phase was used to calculate a fixed value to represent the noncontact phase. A subjective threshold value was chosen based on this average value. Coordinates below this threshold indicated the frames when the foot was in contact with the treadmill, while TO was defined as the frame where the z-coordinates returned above the threshold.

A Bland and Altman method comparison analysis (Bland and Altman 1986) was used to assess the agreement in detecting HS and TO events between the approach described above and visual inspection of the Qualysis motion capture data. The analysis provided 95% limits of agreement, which showed that HS was reliably detected within 0.01 s. There was a consistent discrepancy in TO detection requiring an adjustment of 10 frames to account for this. This discrepancy was attributed to the longer period of unloading that characterises the end of stance (Hausdorff, Ladin, and Wei 1995) and the decreasing influence of the foot on the treadmill marker. Since the loading period of stance was of most interest, the reduced accuracy of TO detection was not considered a serious limitation.

Seven AT subjects presented with unilateral symptoms while 4 had bilateral symptoms of which one leg was randomly selected, resulting in the analysis of 11 injured legs (6 left, 5 right). Angles describing the stance phases of five footfalls for each injured leg and the matched control legs were obtained for BF and shod conditions. The data were time-normalised to 51 data points using MATLAB® (The Mathworks Inc., Natick, MA) and plotted at 2% intervals of total stance time. The angle–time curves were constructed in Excel to allow a visual qualitative analysis. Angular positions at HS, maximum angle, and range of motion (ROM) were calculated for the first 60% of stance. A general linear model split plot ANOVA was carried out on all dependent variables (SPSS v11.0, 2001). This ANOVA model had one between-subject factor (group) and three within-subject factors: condition (2 levels: BF and shod), trial (5 levels), and measure (3 levels: HS, maximum and ROM). The model was run separately for each of the five angles. Alpha was set at $p < 0.1$, and marginal means were plotted with 90% confidence intervals (CIs), as it has been suggested that 95% levels may be too conservative (Batterham and Hopkins 2005). Effect sizes were indicated by partial eta squared (η_p^2) values provided in the SPSS output. Interpretation of η_p^2 was based on Hopkins (2003), where 0.04 to 0.25, 0.26 to 0.58, 0.59 to 0.79, and >0.80 represented small, medium, large, and very large effect sizes, respectively. The condition × group interaction effects were further examined to calculate the magnitude of differences between groups and conditions. Significant pairwise comparisons were determined by nonoverlap of the mean values with the CI bars of another group.

RESULTS

All AT subjects were involved in running or sports where running was a major element. Duration of symptoms ranged from 6 months to 15 years (mean: 43 months ± 51.5 months). Ten out of 11 subjects reported morning stiffness, while 2 reported crepitus. Five subjects had AT pain on walking, while 8 had pain that prevented running. Eight had pain at the start of the run, but this disappeared during the run in 4 cases. Eight subjects reported stretching on greater than 50% of occasions before and after training, but this appeared to have limited preventative action against injury. The podiatric examination revealed little difference in calcaneal alignment between groups during relaxed standing. In the AT group, 7 were everted, 3 vertical and 1 inverted, while in the control group, 6 were everted, 3 vertical and 2 inverted. In all but 1 subject, alignment was the same in right and left legs, regardless of the presence of injury.

Kinematic analysis revealed high between-trial consistency across the five footfalls obtained for each subject in BF and shod conditions (see Figure 1(i) for exemplar data). The average between-trial variation for each subject across all HS, maximum, and ROM measures was 0.95° (range: 0 to 3.59°). In contrast, high between-subject variation was observed in each group, especially in EV angles; see Figure 1(ii) and 1(iii). This variation masks the differences between groups. Figure 2 shows the limited differences observed

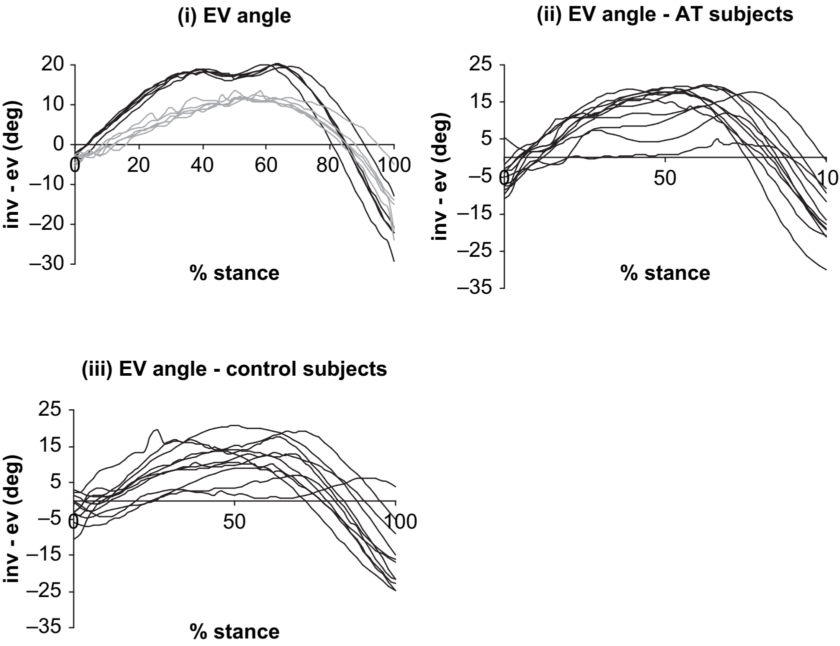

Figure 1. (i) Exemplar data from an AT subject illustrating the high level of consistency across repeated trials in BF (grey lines) and shod (black lines) conditions. Similar repeatability was observed in ADF, leg ABD, calcaneal, and KF angles; (ii) average EV angle-time series curves for AT subjects; and (iii) control subjects. Between-subject variation is high for both groups, making it difficult to distinguish between groups based on qualitative inspection of these curves.

between the mean leg abduction (ABD), calcaneal, EV, ADF, and knee flexion (KF) angle-time series curves obtained for AT and control groups in shod running. The standard deviation curves are included to illustrate that the mean AT group curves fall within the range of the control group curves and vice versa. Table 2 provides the mean HS, maximum, and ROM values calculated for each angle in both running conditions for both groups. The standard deviations indicate similar levels of variation within each group.

Statistical analysis revealed no significant difference between groups for any measure. Several statistically significant differences were found between BF and shod conditions for all subjects regardless of group membership (see Table 3). According to Hopkins' (2003) criteria, there were large effect sizes for maximum KF, maximum EV, and EV ROM. Angular differences exceeding 7° were observed in EV angle measures. Medium effect sizes were calculated for leg ABD, calcaneal, and

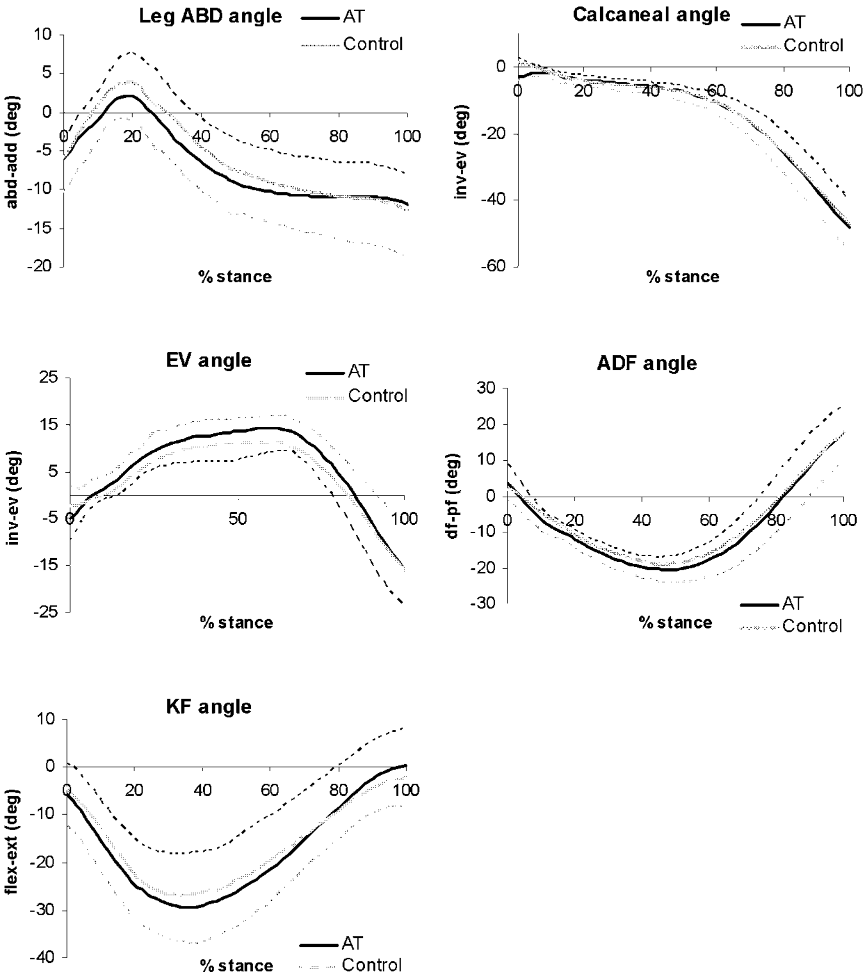

Figure 2. Mean leg ABD, calcaneal, EV, ADF, and KF angle-time series curves for AT and control groups during shod running. The dotted lines represent the standard deviation curves. A high level of between-subject variation was seen in both groups, resulting in overlap of the range of curves for each group.

KF ROM with differences ranging from 1.92° to 2.78°. A significant group × condition interaction effect with a medium effect size was found for maximum KF angle ($p = 0.017$, $\eta_p^2 = 0.309$). Plotting the marginal means to illustrate the interaction effects revealed more detailed information. Significant differences in pairwise comparisons when 90% CIs were plotted are indicated by the relevant symbols in Table 2. The magnitude

Table 2. Means With Standard Deviations in Parentheses of HS, Maximum and ROM Values for Each Angle for AT and Control Groups in BF and Shod Conditions

	BF HS (°)	Shod HS (°)	BF Max (°)	Shod Max (°)	BF ROM (°)	Shod ROM (°)
AT						
Leg ABD	–5.55 (3.05)	–6.08 (2.28)	1.17 (3.78)*	2.64 (4.74)*	6.72 (4.26)*	8.72 (4.48)
Calcaneal	0.82 (3.05)§	–2.94 (6.18)*	0.85 (3.03)	–0.14 (3.91)	0.03 (0.11)	2.79 (3.15)
EV	–2.76 (2.41)	–5.07 (4.62)*	7.85 (2.85)	16.07 (4.29)	10.62 (2.59)	21.14 (5.25)*
ADF	1.11 (5.00)	3.34 (5.46)	–22.32 (4.87)	–21.66 (3.38)	23.42 (3.60)*	25.00 (4.06)*
KF	–3.93 (6.53)	–4.49 (6.76)	–25.47 (8.74)	–29.14 (8.16)	21.54 (4.19)	24.65 (2.54)
Control						
Leg ABD	–3.70 (4.77)	–5.09 (4.34)	5.46 (3.61)	5.92 (3.95)	9.16 (4.21)	11.01 (4.66)
Calcaneal	0.06 (3.21)	0.82 (5.73)	0.10 (3.19)	2.17 (4.18)	0.04 (0.13)§	1.35 (2.35)
EV	–3.83 (3.79)	–2.53 (3.81)	7.73 (4.03)	13.84 (4.93)	11.56 (2.38)§	16.37 (5.11)
ADF	1.50 (4.25)	2.90 (4.88)	–19.80 (3.84)	–19.36 (4.59)	21.29 (3.21)	22.26 (3.18)
KF	–5.70 (5.37)	–4.48 (5.29)	–27.03 (8.22)	–28.03 (7.88)	21.33 (5.02)	23.78 (3.70)

*Indicates differences in pairwise comparisons between AT and control groups when 90% CIs were plotted.

§Indicates differences in pairwise comparisons between BF and shod kinematics within the same group when 90% CIs were plotted.

Table 3. Significant Main Effects for Condition in the Relevant Leg ABD, Calcaneal, EV, and KF Angle Measures. The Angular Differences, *p* Values, and Effect Sizes for the Relevant HS, Maximum, and ROM Measures of Each Angle Are Provided

	Condition main effect		
Angle	**HS**	**Max**	**ROM**
Leg ABD	0.96°, $p = 0.074$, $\eta_p^2 = 0.186$	0.97°, $p = 0.096$, $\eta_p^2 = 0.164$	1.92°, $p < 0.001$, $\eta_p^2 = 0.567$
Calcaneal	–	–	2.04°, $p = 0.007$, $\eta_p^2 = 0.379$
EV	–	7.16°, $p < 0.001$, $\eta_p^2 = 0.697$	7.67°, $p < 0.001$, $\eta_p^2 = 0.644$
KF	–	2.45°, $p < 0.001$, $\eta_p^2 = 0.643$	2.78°, $p < 0.001$, $\eta_p^2 = 0.565$

of these angular differences between groups ranged from 2.13° to 4.77°, with the AT group displaying more pronounced EV and ADF during stance but less ABD of the lower leg compared with the control group. The largest difference was observed in EV ROM during shod running. Results showed that when there were significant differences between shod and BF running, shoes always exaggerated movements, in particular ROM values. Greatest differences were observed in EV measures, with larger differences between BF and shod running in the AT group (8.21°–10.52°) compared with the control group (4.81°–6.11°). This suggests that the shoe exaggerated maximum and ROM values more in the AT group (see Figure 3).

DISCUSSION

All subjects in this study were highly active individuals, suggesting that AT injury is related to activity levels. This supports the majority of previous research findings, but it contrasts with the results of Rolf and Movin (1997) and Åström (1997), who reported the presence of AT injury in inactive individuals (both cited in Alfredson and Lorentzon 2000). The majority of AT subjects indicated that they stretched on at least 50% of occasions before and after training; however, this may still represent a substantial number of occasions where stretching did not take place. McCrory et al. (1999) also found a reduced propensity for stretching in injured AT runners. As the subjects in the current study were not injured at the time of testing, it is possible that many had incorporated stretching into their training routine as part of their rehabilitation. The results of this study and the McCrory et al. study are limited, as subjects were asked to

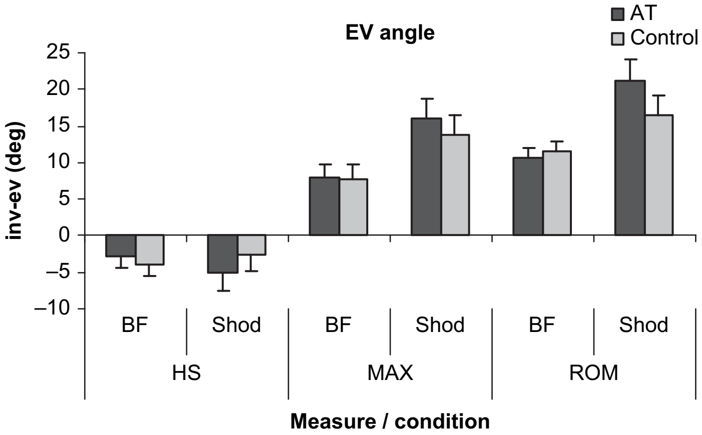

Figure 3. Mean EV angle measures plotted with 90% CIs in BF and shod running conditions for AT and control groups. Maximum and ROM measures are exaggerated in the shod condition for both groups, but particularly in the AT group.

provide information about the areas stretched (hamstrings, quadriceps, and calves) but the duration, type, and quality of stretching was not assessed.

Kinematic Differences Between AT and Control Groups

Pronation is commonly is associated with running injuries, but a definite link has not been established (Payne 1999). Literature has reported higher levels of pronation in clinical populations such as those with Achilles tendonitis (McCrory et al. 1999) and shinsplints (Messier and Pittala 1988; Viitasalo and Kvist 1983) compared with controls. The AT subjects in this study presented high levels of pronation during running as determined by the podiatrist in their initial consultations. While some pronation is necessary for shock absorption and effective movement, researchers have been unable to define the level beyond which it becomes excessive, as this appears to be subject specific (Hamill, Haddad, and van Emmerik 2006). Undesirable levels combined with structural or functional misalignments and repetitive activities such as running are thought to be risk factors for injury.

Substantial efforts were made in this study to recruit subjects with a specific injury and similar kinematic patterns to standardize the mechanism of injury. Although numerous other factors are involved, it was suggested that this subject control would induce more systematic movement patterns. As subjects from both groups were well matched in static alignments, any kinematic differences may be attributed to the mechanisms resulting in injury. Results showed high between-subject variation but a high level of consistency in within-subject kinematics. This resulted in small differences between the means, which may have masked some differences between groups. In BF and shod running, the AT group displayed less maximum leg ABD and ROM but greater ADF ROM. Additionally, the AT group displayed greater inversion (INV) at HS and greater maximum EV and EV ROM during stance. This supports the two-dimensional findings of McCrory et al. (1999), who found greater INV at HS and increased EV-related measures during stance in Achilles tendonitis subjects. While McCrory et al. found differences of <0.7° in maximum EV and EV ROM between groups, the current study found substantial differences of 2.23° and 4.77° in the same measures. McClay and Manal (1998) also found greater EV and ADF measures in excessive pronators compared with controls.

Reduced ankle flexibility due to tightness in the gastrocnemius or soleus has been linked to AT injury (Clement et al. 1984; Cook et al. 2002; Hutson 1996; Kaufman et al. 1999). This has been attributed to prolonged contraction of the gastroc-soleus complex in an attempt to control pronation (Nicolopoulos, Scott, and Giannoudis 2000). The AT subjects in this study were diagnosed with an ankle equinus, indicating limited passive ADF. Theoretically, this reduced movement could increase the strain on the AT during running (Clement et al. 1984), but this is contradicted by the trend for greater maximum ADF in the AT group compared with controls during running. Alternatively, increased ADF may have been due to weakness in the plantar flexor muscles, or as a result of prolonged foot contact on an unstable surface (Hutson 1996). Research has found longer stance periods on a treadmill, which may be classified as an unstable surface, but as conditions were identical for all subjects, this should not be a major factor between groups.

Kinematic Differences Between BF and Shod Running

Injured subjects would be expected to present undesirable movement patterns either when running BF or in shoes. Stacoff et al. (2000) found small, unsystematic differences in EV between BF and shod running when using bone markers in uninjured subjects. As external marker movement artefact typically overestimates frontal and sagittal plane motion in shod running (Reinschmidt et al. 1997), some differences were expected in this study. Shod running generally exaggerated EV and KF kinematics, resulting in greater movement. Most discrete differences between conditions were <4°, except for the EV angle where differences of 8.21°–10.52° were observed for the AT group. Notably, lower differences of 4.81°–6.11° were observed for the control group (see Table 3). This indicated that the shoe exaggerated the kinematics to a greater extent in the AT group than in the control group. The shoe failed to provide adequate movement control in this group, which may be of particular importance in injury occurrence.

While BF running provides a better indication of the true motion of the foot, it does not reflect how it behaves with the addition of the primary external device, the shoe. Research has indicated that injury is more likely to occur in shod running than BF running (Robbins and Hanna, 1987, cited in Warburton 2001). If excessive EV is linked to injury, this is logical given the greater EV in shod running. The greater ROM may partly explain the 3%–5% increase in energy expenditure observed in this condition compared with BF running (Nigg and Segesser 1992). This greater energy cost coupled with the repetitive motions of running may increase the likelihood of developing an overuse injury such as an AT injury. While subjects wore their own shoes for this study, they were all running shoes with similar design features. Previous research has found kinematic differences induced by shoe modifications to be small and unsystematic (Stacoff et al. 2001), suggesting that this factor may have limited effects. Unsuitable design features such as a badly fitting or too soft heel counter, inadequate heel wedging, and an inflexible sole may allow excessive movement and increase strain on the AT (Smart et al. 1980). The results suggest that footwear selection may be important for AT subjects, as they seem unable to control the range of movement compared with the controls. This results in higher levels of movement that may be associated with injury.

LIMITATIONS

Kinematic gait studies typically have a number of design limitations that are difficult to resolve. External markers cannot directly measure talar movement, instead providing information on the combined motion of the subtalar and talocrural joints (McClay and Manal 1998). Shoe markers represent shoe motion rather than the underlying bone motion (Edington, Frederick and Cavanagh 1990; Reinschmidt et al. 1997; Stacoff et al. 2000), and hence do not indicate true movement of the foot. While the marker set-up used in this study provided highly repeatable frontal and sagittal plane angle data, previous work found unacceptable levels of variation in transverse plane motion. Foot abduction and tibial rotation have been coupled to motion in the other planes, and it has been suggested that they may be important factors in the occurrence of injury (Areblad et al. 1990; Nawoczenski, Cook, and Saltzman 1995). Examination of transverse plane motion would provide a more complete analysis of foot and lower limb motion. Lower limb kinematics typically are measured relative to a reference posture to account for the different magnitude of motion in individual movement patterns. The method of obtaining subtalar joint neutral position is an accepted and integral aspect of clinical podiatric examination. The validity of obtaining this position has been debated as intertester and between-test reliability is quite low (Menz 1998). It is unreasonable to consider that differences of 4°–11° between groups and conditions could be attributed solely to inaccuracy in measuring subtalar joint neutral position. Since all measurements were expressed relative to this position, ROM measures would not be affected, reinforcing the validity of these values, in which many differences were observed.

CONCLUSION

The data revealed qualitative differences in the angle–time curves between AT and control groups in BF and shod running. The results showed high levels of within-subject consistency in all measures, but between-subject variation was high in both AT and control groups. There were clear distinctions between groups, with AT subjects displaying greater EV and ADF but less leg ABD during stance compared with controls. Shod running exaggerated ROM values in three angles, particularly the EV angles for the AT group. This study examined discrete measures during stance using a traditional statistical approach. This provided limited information, as it did not reflect the entire sequence of movement during foot contact and could not cope with the inherent variability of individual movement patterns. Curve-based approaches such as functional data analysis is recommended for further analysis of these data. This approach would analyze the entire time series data as a function rather than as a series of discrete parameters (Ryan, Harrison, and Hayes 2006). Future analysis also should examine the kinematics of subjects presenting with eccentric lengthening of the gastrosoleus complex during stance to compare subjects with the same injury but different mechanisms of injury.

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