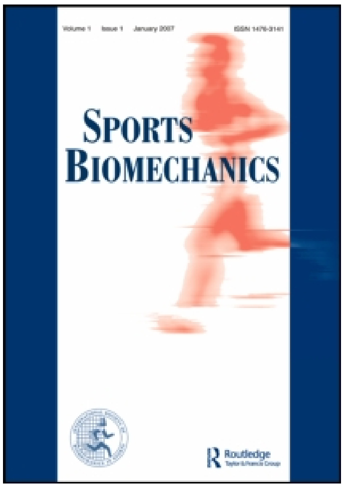

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Orthotic control of rear foot and lower limb motion during running in participants with chronic Achilles tendon injury

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Abstract
This study examined the kinematic effects of orthoses in participants with a history of chronic Achilles tendon injury. Twelve participants ran at self-selected speeds on a treadmill with and without customized orthoses. Joint and segment angles including leg abduction, calcaneal, eversion, ankle dorsiflexion, and knee flexion angles were calculated from three-dimensional data throughout stance. Five footfalls were obtained for each participant and condition. Statistical tests revealed an increase in maximum eversion with orthoses ($P < 0.001$, $\eta_p^2 = 0.642$). In the individual participant analysis, this was evident in 9 of 12 participants. Trends towards increased eversion range of motion and decreased ankle dorsiflexion maximum and range of motion angles were also observed. Increased eversion was unexpected as all devices were designed to provide pronation control as deemed necessary by the podiatrist. Despite this, participants reported between 50 and 100% (average 92%) relief from symptoms with the use of orthoses. Further analysis of the angle–time curves and coordination between angular measures is recommended.

Keywords: *Eversion, in-shoe devices, kinematics, lower limb*

Introduction

Orthoses are commonly used in the treatment of lower limb pathologies, including patello-femoral pain, shin splints, and Achilles tendon injury. Reviews of the literature conclude that these devices often provide symptom relief for these injuries, but they do so through mechanisms that are not well understood (Ball and Afheldt, 2002; Heiderscheit, Hamill, and Tiberio, 2001; Kilmartin and Wallace, 1994; Landorf and Keenan, 1998; Pratt, 2000). It is thought that orthoses may have effects on joint kinematics, plantar pressure, muscle activity, oxygen consumption, pain, deformity, and proprioception (Ball and Afheldt, 2002; Landorf and Keenan, 1998). Much of the literature has focused on orthotic control of foot and lower limb movements (Eng and Pierrynowski, 1994; Ferber, Davis, and Williams, 2005;

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Johanson, Donatelli, Wooden, Andrew, and Cummings, 1994; Stacoff et al., 2000), but the results indicate small, subtle, and inconsistent changes, making it difficult to ascertain exactly how orthoses work.

Some inconsistencies in the results of previous research may be due to limitations in experimental design. These limitations include differences in injuries or symptoms presented by the participants under analysis, the examining and manufacturing techniques used by clinicians, and the structure and design of the orthoses. Few studies to date have examined a specific injury (Eng and Pierrynowski, 1994), but many have focused on common factors such as forefoot varus and excessive pronation (Johanson et al., 1994; McCulloch, Brunt, and Vander Linden, 1993; Mundermann, Nigg, Humble, and Stefanyshyn, 2003). These functional alignments may be common to a range of injuries but as each condition tends to be multifactorial, the mechanisms of injury are often injury-specific. Ideally, a group with the same movement pattern and same mechanism of injury should be recruited so that the prescribed orthoses are based on the same design principle. Groups of uninjured participants with no history of malalignment or orthotic use have also been analysed (Nester, van der Linden, and Bowker, 2003; Stackhouse, Davis, and Hamill, 2004; Stacoff et al., 2000), although Heiderscheit et al. (2001) questioned the value of examining orthotic effects in these individuals, as they may respond differently to injured participants. Some studies provided individuals with custom-made orthoses (Bates, Osternig, Mason, and James, 1979; Eng and Pierrynowski, 1994; Ferber et al., 2005; McCulloch et al., 1993), whereas others provided a standard wedge or post to all devices (Mundermann et al., 2003; Nester et al., 2003; Stackhouse et al., 2004; Stacoff et al., 2000). Orthotic materials vary across studies with soft (Eng and Pierrynowski, 1994), semi-rigid (McCulloch et al., 1993; Nawoczenski, Cook, and Saltzman, 1995), and rigid (Bates et al., 1979; McCulloch et al., 1993) devices being used. Consequently, the structural and functional factors of the individual have not always been taken into account. For effective analysis of orthotic effects, standardization of the diagnostic and production techniques is necessary.

Chronic Achilles tendon injury affects many of those involved in running and jumping activities and, in extreme cases, can prevent exercise participation. The condition does not always respond well to physiotherapy-based treatments as symptom relief is usually temporary. Eccentric exercises appear to provide the most effective pain resolution (Alfredson, Pietila, Jonsson, and Lorentzon, 1998), while anecdotal evidence also supports the use of orthoses. Research that examines how a specific orthotic construction will control a particular mechanism of injury is lacking. Controlling the mechanism and resultant injury provides a standardized function for the orthoses. The aim of this study was to examine the effect of customized orthoses on lower limb kinematics in participants with a history of chronic Achilles tendon injury. Since the mechanism of injury in these participants was thought to involve torsional stress due to high pronation (Clement, Taunton, and Smart, 1984), it was hypothesized that orthoses would reduce pronation-related measures such as ankle dorsiflexion and eversion during stance compared with when orthoses were not worn.

Methods

Participants

Ethical approval for this study was obtained from the university ethics committee. Based on *a priori* power analysis (Lenth, 2005), 12 participants was considered to be sufficient to detect an angular difference of 1.5° ($\alpha = 0.05$, $1 - \beta = 0.997$). The Podiatry Department private patient files at the university were examined to identify patients who presented to the collaborating podiatrist in the year before testing with Achilles tendon pain. Based on patient files, all those displaying amounts of pronation during running that, based on the podiatrist's judgement, were likely to be related to the clinical presentation of Achilles tendon injury were invited to participate in the study. Participants with Achilles tendon injury but who displayed a rigid foot type with little visible pronation during running were excluded.

Twelve participants with a history of chronic, low-grade Achilles tendon injury were recruited (age 38.7 ± 8.1 years, height 1.75 ± 0.05 m, body mass 73.3 ± 8.5 kg; mean ± *s*). All participants had already undergone a podiatric assessment, which consisted of a series of clinical observations. This involved a subjective, qualitative assessment of the alignment of the foot, amount of inversion–eversion and rigidity–laxity of the foot during barefoot running, and calcaneal alignment in the relaxed standing position and supine subtalar neutral position. The podiatrist had provided all participants with customized orthoses that were constructed on a neutral cast of the individual's foot. They were made of high-density ethylene vinyl acetate, which was heat moulded and machined to fit the participant's foot and shoes and provide appropriate control as diagnosed by the podiatrist. Of four participants with bilateral injuries, one leg was randomly selected (by coin toss) for analysis resulting in 12 injured legs of which 6 were left and 6 were right. All participants continued to wear their orthoses and were asymptomatic at the time of testing.

Experimental set-up and testing

The procedures carried out during the initial podiatric consultation as outlined above were repeated by a second collaborating podiatrist before testing. There was good agreement between the qualitative assessments of both clinicians. Participants completed a questionnaire retrospectively, providing a history and description of the injury, information about training, treatments received, and any subsequent improvements. During running trials, all participants wore their own running shoes. This was required since custom-made orthoses are designed to fit the individual's own shoe and this study sought to provide a realistic interpretation of how orthoses affect their normal movement patterns. As this study was concerned with the relative differences between orthoses and no-orthoses conditions (repeated-measures design), the shoe represents a constant factor between conditions and, therefore, would have limited influence on the results. A marker set-up similar to that used by Clarke et al. (1983) was used. Retro-reflective markers were placed on the posterior and lateral aspects of both lower extremities as follows: two on the posterior aspect of the shoe bisecting the heel, two bisecting the posterior shank (one on the Achilles tendon, one below the belly of the gastrocnemius), one on each of the fifth metatarsals (fixed to the surface of the shoe), lateral malleoli, fibular heads, and greater trochanters. All participants were experienced in treadmill running, but completed a familiarization session before data capture. This involved continuous running on the treadmill at various speeds for a minimum of 4 min. Previous work found high consistency in the lower limb kinematic parameters of interest across repeated trials when using a treadmill (Donoghue and Harrison, 2004). This consistency allowed kinematic changes to be attributed to orthotic effects rather than random trial-to-trial variations.

Three-dimensional kinematic data were captured using Qualysis Motion Capture Systems (Gothenburg, Sweden) and eight synchronous ProReflex MCU240 cameras, operating at a sampling frequency of 200 Hz. Cameras were located in an arc around the posterior and lateral aspects of the treadmill. A marker was placed on the rigid frame at the front of the treadmill to indicate the position of the treadmill surface on the motion capture screen. The podiatrist placed all participants in the subtalar neutral stance position before each running condition and the marker coordinates were obtained. Participants ran at the same self-selected speed (2.8 ± 0.3 m/s) in the orthoses and no-orthoses conditions. Data capture took place during the third minute of continuous running and each participant indicated when they had fully recovered (minimum 4 min) before completing the next condition. Previous research has found limited speed-related effects on rear foot kinematics (Donoghue and Harrison, 2004), suggesting that any variation in self-selected speeds should have minimal effects on the joint kinematic parameters measured in this investigation.

Data analysis

Raw marker coordinate data were exported from Qualysis and imported as scaled coordinates into the Peak Motus^{TM} Analysis System (Peak Performance Technologies, Englewood, CO, USA). The frontal and sagittal plane angles described in Table I were calculated from three-dimensional kinematic data. Eversion and ankle dorsiflexion were considered to be of most interest as they are components of pronation, which has been linked specifically to Achilles tendon problems (Clement et al., 1984). High amounts of pronation were also a functional characteristic of these participants. Data were exported to Microsoft Excel for further analysis. All angles were calculated relative to the subtalar joint neutral position taken before the dynamic trials. Heel strike and toe-off 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 non-contact 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 toe-off 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 compare the agreement in detecting heel strike and toe-off 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 heel strike was reliably detected within 0.01 s. There was a consistent discrepancy in toe-off detection requiring an adjustment of 10 frames to account for this. This discrepancy was attributed to the longer period of unloading that characterizes the end of stance (Hausdorff, Ladin, and Wei, 1995) and the decreasing

Table I. Definitions of angles calculated to describe rear foot and lower leg kinematics.

Angle	Definition
Leg abduction angle	Angle between the lower leg and the ground on the medial side as viewed from behind; indicates abduction–adduction of lower leg
Calcaneal angle	Angle between the rearfoot and the ground on the medial side as viewed from behind; indicates inversion–eversion of rearfoot
Eversion angle	Segment angle between rearfoot and lower leg; indicates inversion–eversion of rearfoot relative to the lower leg
Ankle dorsiflexion angle	Anatomical joint angle between fibular head, ankle, and fifth metatarsal; indicates ankle dorsiflexion–plantar flexion
Knee flexion angle	Anatomical joint angle between greater trochanter, fibular head, and ankle; indicates knee flexion–extension

influence of the foot on the treadmill marker. Since orthoses are usually designed to influence the foot during early stance (Nawoczenski et al., 1995), the limited accuracy of toe-off detection was not considered a serious limitation.

The stance phases of five footfalls for each leg and for each condition were obtained. The data were time-normalized to 51 data points using MATLAB® (The Mathworks Inc., MA, USA) and data were plotted at 2% intervals of total stance time. The angle–time curves were constructed in Excel to allow a visual qualitative analysis. Angular values at heel strike, maximum–minimum angular displacement (depending on direction of time series curve), and range of motion were calculated for the first 60% of stance. Statistical tests (SPSS v11.0, 2001) included a general linear model repeated-measures analysis of variance (ANOVA) for each of the five angles with two factors included in the model: condition with two levels (orthoses and no-orthoses conditions) and trial with five levels. The three dependent variables were heel strike, maximum and range of motion measures for each angle. Probability values were calculated and marginal means were plotted with 90% confidence intervals in accordance with the recommendations of Sterne and Smith (2001). Effect sizes were indicated by the partial eta squared (η_p^2) values provided in the SPSS output. Interpretation of η_p^2 was based on the scale used by Comyns et al. (2007), where 0.04–0.249, 0.25–0.549, 0.55–0.799, and >0.80 represented small, medium, large, and very large effect sizes, respectively. Criteria were also determined to indicate a substantial change in individual joint kinematics induced by orthoses. The procedure for this involved calculating the within-participant standard deviations across five trials for heel strike, maximum and range of motion measures in orthoses and no-orthoses conditions. The average standard deviation across all of these measures and across all participants was then calculated (range 0.65–1.05°). These values were regarded as indicators of the variation observed across repeated trials; when multiplied by 1.645, the 90% confidence interval (CI) for this variation was obtained. An angular difference between conditions that exceeded these criteria was deemed to be substantial enough to be of practical significance. Participant responses to orthoses were then classified based on a substantial increase, decrease or no change in these discrete kinematic values.

Results

Questionnaire and observation results

All participants were involved in running or sports where running was a major element. The majority reported morning stiffness and pain either at the start of the run or pain that prevented running. Duration of symptoms ranged from 6 months to 15 years (mean 42 ± 48 months). Seven participants indicated that they sought advice from physiotherapists, experiencing varying levels of temporary improvement; the mean reported pain relief was 58 ± 38% (range 0–90%). Ten participants reported a mean improvement of 92 ± 16% (range 50–100%) when wearing orthoses, with five participants indicating complete symptom resolution. Of the two participants who did not respond, one had just received orthoses and may not have had sufficient time to adjust to them.

Kinematic results

Table II shows the means and standard deviations of heel strike, maximum and range of motion measures for each angle in orthoses and no-orthoses conditions. Similar values were observed in both conditions for most measures; differences exceeding 2° were only seen

Table II. Heel strike (HS), maximum (Max.) and range of motion (ROM) measures for leg abduction (ABD), calcaneal, eversion, ankle dorsiflexion (DF), and knee flexion angles in orthoses (O) and no-orthoses (NO) conditions (mean $\pm s$).

Angle	O HS (°)	NO HS (°)	O Max. (°)	NO Max. (°)	O ROM (°)	NO ROM (°)
Leg ABD	−5.2 ± 3.9	−5.6 ± 2.5	3.4 ± 4.4	3.3 ± 4.5	8.6 ± 5.0	8.9 ± 4.8
Calcaneal	−2.3 ± 5.3	−2.8 ± 5.6	−0.1 ± 3.1	−0.5 ± 3.6	2.2 ± 2.8	2.3 ± 2.7
Eversion	−2.8 ± 4.0*	−4.3 ± 4.3	19.3 ± 4.5*	16.5 ± 4.0	22.0 ± 4.7*	20.7 ± 4.7
Ankle DF	2.2 ± 4.4	2.4 ± 5.5	−20.5 ± 3.7	−21.1 ± 3.3	22.6 ± 5.6	23.5 ± 5.5
Knee flexion	−3.9 ± 7.3	−6.0 ± 6.3	−27.6 ± 9.1	−27.4 ± 8.8	23.7 ± 4.5	23.2 ± 4.5

* Differences between orthoses and no-orthoses conditions with $P < 0.01$ and medium to large effect sizes.

in knee flexion angle at heel strike and maximum eversion. Probability values and effect sizes indicated that condition discriminated between eversion angle at heel strike (difference = 1.48°, $P = 0.046$, $\eta_p^2 = 0.315$), maximum eversion (difference = 2.79°, $P < 0.001$, $\eta_p^2 = 0.642$), and eversion range of motion (difference = 1.30°, $P = 0.067$, $\eta_p^2 = 0.273$). According to the criteria of Comyns et al. (2007), these indicate a large effect size for maximum angle and medium effect sizes for heel strike and range of motion angles. All remaining measures had P-values > 0.1 and small effect sizes indicating that they were poor discriminators between orthoses and no-orthoses conditions. Plotting the mean values for all measures and angles with 90% CI bars supported this increase in maximum eversion with the use of orthoses (see Figure 1).

Figure 2 shows exemplar angle–time curves for an individual in orthoses and no-orthoses conditions. While the general patterns of the curves are consistent, the graphs revealed subtle but clearly identifiable kinematic differences when running with and without orthoses. This clear distinction between conditions is representative of the kinematics observed in all participants. Since the group means revealed limited differences between conditions, means and standard deviations for heel strike, maximum and range of motion values of each angle (across five trials for each participant) were also examined. The within-participant variation between trials was very low, often less than 1°. The between-participant variation in positions obtained during stance and responses to orthoses was substantially higher. Figure 3 illustrates this for all ankle dorsiflexion and eversion angle measures in orthoses and no-orthoses conditions.

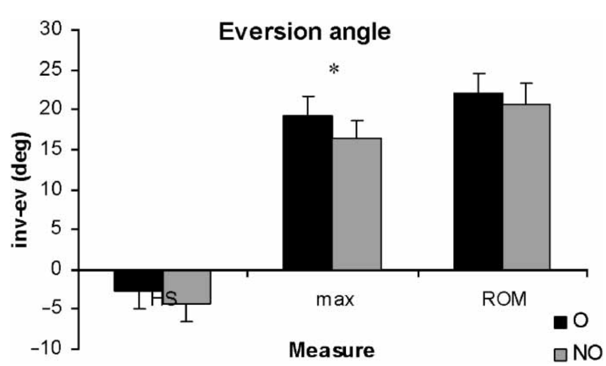

Figure 1. Marginal means for eversion angle measures (HS = heel strike, max = maximum, range of movement = range of motion) plotted with 90% CI bars (inv-ev = inversion–eversion). * Non-overlap of the 90% CI bars for maximum eversion angle in orthoses (O) and no-orthoses (NO) conditions.

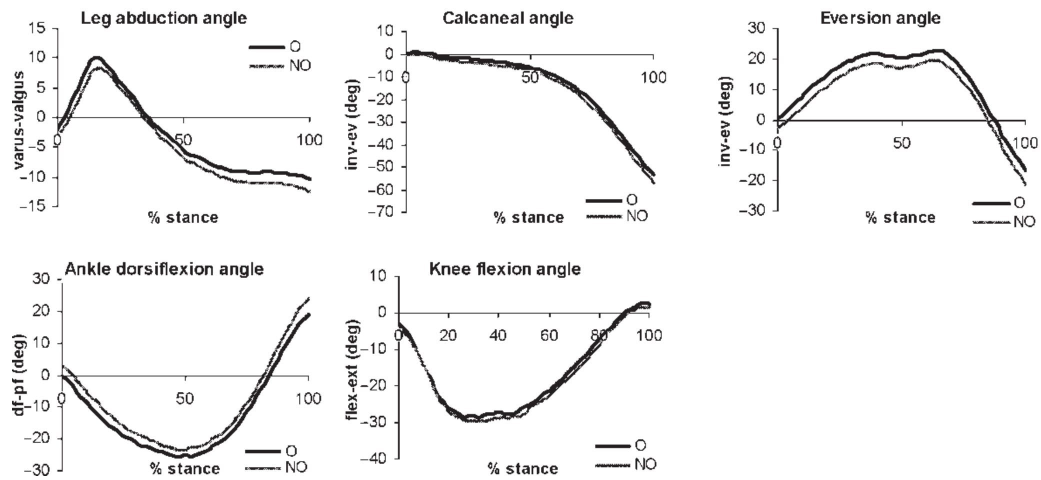

Figure 2. Exemplar angle–time curves for a participant in orthoses (O) and no-orthoses (NO) conditions. These curves represent the symptomatic leg and show small but distinct differences between the means for each condition (df-pf = dorsiflexion–plantar flexion, inv-ev = inversion–eversion, flex-ext = flexion–extension).

Subsequently, the magnitude and consistency of the effects of orthoses on each participant's movement pattern was examined. Table III shows the number of participants who responded with a substantial increase or decrease or no change in heel strike, maximum and range of motion measures for each angle with orthoses. Some trends did emerge with orthoses promoting a more everted rearfoot position throughout stance and greater maximum eversion in 9 of 12 legs. One-third of participants showed a decrease in maximum

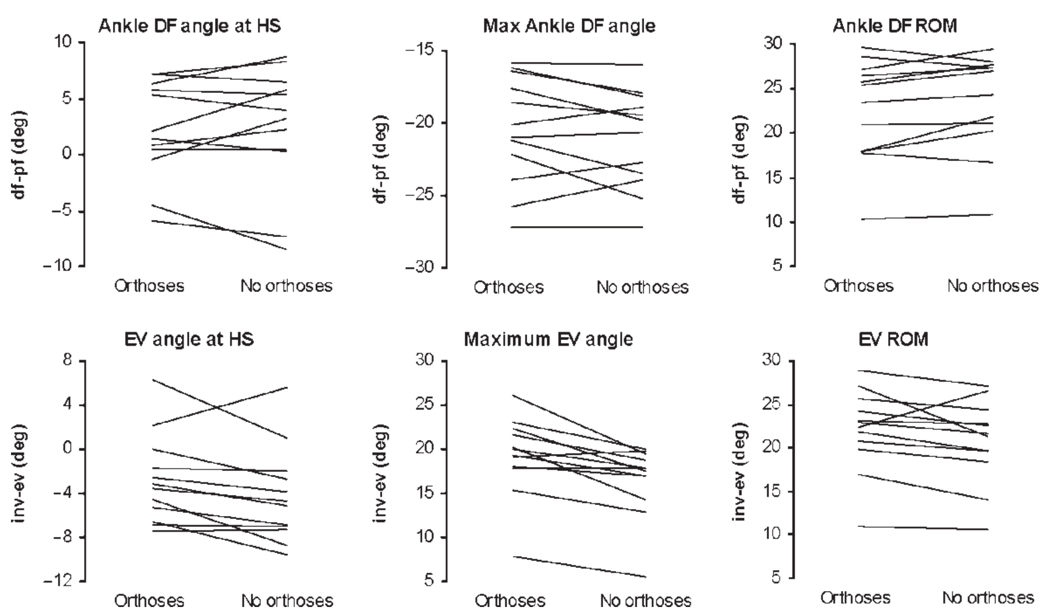

Figure 3. Average heel strike (HS), maximum and range of motion (range of movement) measures for ankle dorsiflexion (DF) and eversion (EV) angles for individual participants in orthoses and no-orthoses conditions (df-pf = dorsiflexion–plantar flexion, inv-ev = inversion–eversion).

Table III. Number of participants displaying substantial changes in heel strike, maximum and range of motion values for each angle with the use of orthoses ($n = 12$).

		Number of participants		
Effect of orthoses	Average within-participant $s \times 1.645$	Increase	Decrease	No change
Leg ABD at HS (↑ = ↑ varus)	1.48°	4	4	4
Leg ABD max. (↑ = ↑ valgus)		3	2	7
Leg ABD range of motion		0	3	9
Calcaneal at HS (↑ = ↑ INV)	1.11°	1	3	8
Calcaneal max. (↑ = ↑ EV)		2	1	9
Calcaneal range of motion		1	2	9
Eversion at HS (↑ = ↑ INV)	1.69°	1	5	6
Eversion max. (↑ = ↑ EV)		9	0	3
Eversion range of motion		5	1	6
Ankle DF at HS (↑ = ↑ DF)	1.60°	3	2	8
Ankle DF max. (↑ = ↑ DF)		1	4	7
Ankle DF range of motion		0	4	8
KF at HS (↑ = ↑ flexion)	1.68°	4	3	5
KF max. (↑ = ↑ flexion)		4	3	5
KF range of motion		2	0	10

Note: Substantial changes were indicated by average within-participant standard deviation (s) × 1.645; see text for further explanation of how these were calculated (HS = heel strike, max. = maximum, ABD = abduction, DF = dorsiflexion, KF = knee flexion).

ankle dorsiflexion and ankle dorsiflexion range of motion when wearing orthoses with the remainder showing no change. Participants showed predominantly no change or relatively equal variation in responses to orthoses for the remaining angles.

Discussion and implications

Clinicians perceive excessive pronation and knee flexion during locomotion as abnormal. Pronation is the factor that is most commonly associated with running injuries, although a causative link has yet to be established (Payne, 1999). Literature has reported the presence of greater pronation in clinical populations such as those with Achilles tendonitis (McCrory et al., 1999) compared with controls. While pronation provides necessary shock absorption during stance and allows for effective locomotion, it can be detrimental if too large and prolonged (Clement et al., 1984). Undesirably large pronation, combined with structural or functional misalignments and repetitive activities such are running, are thought to be a risk factor for injury. However, the level of pronation that is deemed excessive is not yet clear.

The results of this study showed strong statistical support for increased eversion throughout stance when orthoses were worn, in particular maximum eversion. Individual participant analysis supported this, revealing that maximum eversion increased in 75% of cases and range of motion increased in 42% of cases. Changes in eversion kinematics were typically 2–3°, although differences of up to 7° were observed in four participants. This analysis also revealed that ankle dorsiflexion maximum and range of motion angles were reduced in 33% of cases when wearing orthoses (no change in 67%). Reductions of up to 4° were seen in ankle dorsiflexion range of motion. To date, orthoses research has been unable to quantify how much movement control is of practical and clinical significance. It is likely to be participant-specific, depending on the initial kinematics, structural alignment, and presenting symptoms.

The main finding of this study was the increased eversion relative to the lower leg observed during the orthoses condition. Each orthosis consisted of an arch support and a medial wedge and was designed to reduce pronation during ground contact. Therefore, this increase was unexpected as the physical presence of the medial wedge alone should provide a "block" to frontal plane movement. Although inconsistent and unsystematic results are widespread in orthoses literature, similar devices to those used in this study have been found to reduce pronation, especially in the frontal plane (Eng and Pierrynowski, 1994; Stackhouse et al., 2004). Interestingly, there were few substantial differences in calcaneal eversion between the orthoses and no-orthoses conditions, illustrating the small orthotic effects on heel movement. If pronation is associated with injury, it is reasonable to assume that these participants might display new symptoms or no resolution of current symptoms when wearing orthoses. However, 10 of 12 participants reported improvements in their pain symptoms ranging from 50% to 100% regardless of the kinematic changes. It is also noteworthy that other treatments (which experienced some initial success) had difficulty in resolving the long-lasting effects of this condition, which continued for up to 15 years in one case. The link between pronation and injury has been questioned previously as participants have displayed "excessive" pronation but no signs of injury (Payne, 1999).

As ankle dorsiflexion is a component of pronation, it would also be expected to decrease when wearing orthoses. Many participants showed no change but those who did invariably showed a decrease in ankle dorsiflexion during stance. This is possibly due to the elevated heel in the orthoses and the bony architecture of the foot, which provides a natural limit to the ankle dorsiflexion achievable. This supports the findings of Harrison et al. (2001), who reported that a triplanar wedge reduced ankle dorsiflexion range of motion in participants with a history of Achilles tendonitis, but it contrasts with other reports of non-significant increases in ankle dorsiflexion range of motion with orthoses (Stackhouse et al., 2004). As the latter study used uninjured participants and provided mean group data not individual data, the comparison between results is questionable.

As injuries are typically multi-factorial and the mechanisms of each injury differ, a significant amount of control is required to examine similar conditions across participants. This study recruited participants with a specific injury associated with high levels of pronation during stance in an effort to standardize the mechanism of injury. Customized orthoses, designed according to the same principles of pronation control, were then provided to all participants. The same podiatrist carried out the assessment, diagnosis, and manufacturing procedures to ensure consistency in the devices used. While these procedures are largely based on the clinical skills of the podiatrist, it does assess the effectiveness of orthoses in a realistic patient population. It was suggested that this control should induce systematic orthotic effects and that the lack of control over participant recruitment and the use of generic devices may have contributed to the inconsistent results of previous studies. Despite this control, the results showed high between-participant variation (range 2–9°) in discrete measures. This variability was also seen in previous studies (Stackhouse et al., 2004; Stacoff et al., 2000). Smith et al. (2001) suggested that the high individual variability in ankle joint and foot motion in response to any intervention may be explained by the individual interaction between the sensory apparatus and the contacting surface.

In contrast, low within-participant variation (mean 0.92°; range 0.04–2.27°) was found for discrete measures across five trials and for graphical angle–time curves. This high repeatability in successive trials allowed distinct differences between average orthoses and no-orthoses conditions for each participant to be revealed. These small and subtle differences varied between participants and were often masked by the group analysis, but this supports the fact that orthoses did affect the lower leg and foot kinematics. This highlights the limitations of group analysis and suggests that a single participant analysis may be more appropriate. This study calculated a substantial change, taking measurement error and individual variation across repeated trials into account. With the use of orthoses as the sole factor that differed between conditions, any change above the normal trial-to-trial variation was attributed to the use of this device.

This study focused on the kinematic changes induced by orthoses. Discrete measures such as heel strike angle, maximum–minimum angular position, and total range of motion during stance were examined using a multivariate statistical approach. This traditional approach could not explain how these devices effectively relieved the symptoms of injury while increasing frontal plane movement, although these effects may have been specific to this particular injury group. One possibility is that orthoses may have induced foot abduction, which would have increased the tendency for the arch to collapse as the weight progressed forward, thus accounting for the increased eversion. This is due to alterations in the orientation of the ankle and subtalar joint axes relative to the plane of progression. Greatest motion occurs about the axis that is closest to perpendicular to the plane, in this case the subtalar axis (Nordin and Frankel, 2001), in this case the subtalar axis. Since transverse plane motion was not analysed in this study, this hypothesis cannot be confirmed. It is also possible that examining velocity of movement or changes in loading patterns could have provided additional information. While the results showed that the movement patterns changed, the mechanisms involved remain unclear.

Discrete measures provide a limited interpretation of orthotic effects as the entire sequence of lower limb movements during stance is discarded, wasting potentially useful information. It was unclear why frontal plane motion increased but sagittal plane motion decreased, although it may be related to vibrations of the marker placed just below the belly of the gastrocnemius. Examination of the Qualysis motion capture files revealed that immediately after foot impact, this marker oscillated with a sinusoidal pattern. This may have affected the kinematics in an unpredictable but potentially important way, but this study does not have sufficient data to ascertain if vibration plays a role or how it may relate to the observed findings. Further work using high-frequency data capture and analysis of soft tissue movement rather than marker movement would be required.

The inherent between-participant variability in kinematic values provided another limitation to this traditional ANOVA-based approach. This suggests that an alternative approach using curve analysis may be more suitable to extract the relevant information from the data. Functional data analysis is a versatile, statistical technique that examines patterns within curve-based data. It views the entire time series data as a function rather than a series of discrete parameters and has been used to analyse kinematic vertical jump data (Ryan, Harrison, and Hayes, 2006). It can be used for time series and coordination data and it is suggested that this could provide useful information in the future analysis of such data.

Movement coordination has received less attention in the literature than the component movements. In recent years, dynamical systems theory has emerged as a possible explanation for the mechanisms of injury. It uses continuous relative phase approaches to examine movement coordination and variability relationships across stance. Hamill et al. (1999) suggested that injured individuals freeze the degrees of freedom to minimize pain, resulting in the same movement pattern or a reduced number of potential movement patterns. This reduces trial-to-trial variation, which may lead to tissue overload and subsequent injury. Coordination between eversion, ankle dorsiflexion, and knee flexion may be crucial to understanding the mechanisms of injury and the effects of interventions such as orthoses. This should be examined in future research, as the results suggest a possible link between increased frontal and decreased sagittal plane motion.

Conclusion

Significant control was incorporated into this study design, resulting in high within-participant consistency in all kinematic measures. However, between-participant variation in kinematic responses to orthoses remained quite high. Maximum eversion increased when orthoses were worn but there was a trend for decreased maximum ankle dorsiflexion and ankle dorsiflexion range of motion. Despite this, 10 of 12 participants reported substantial improvements (50–100%) in pain relief (average 92.4%). Visual examination of individual angle–time curves revealed evidence of subtle kinematic differences between conditions; however, the magnitude and direction of these effects varied across participants. The exact mechanisms by which orthoses work were not revealed based on this traditional multivariate analysis of kinematic data. Future analysis of such data using a statistical curve-based approach and from a coordination perspective is recommended.

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