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The effect of simple insoles on three-dimensional foot motion during normal walking

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Abstract

Background. The purpose of this study was to establish the effect of simple non moulded flat based insoles on three-dimensional foot motion during normal walking. Excessive foot pronation is considered a major contributing factor to lower limb injuries. Moulded foot orthoses have been shown to decrease maximum foot eversion. Simple insoles are widely used in clinical practice as an alternative to moulded orthoses. However, there has been little research into the kinematic effects of simple insoles.

Methods. All subjects had an inverted rearfoot and forefoot position when the subtalar joint was placed in neutral, which was assessed by a weight bearing goniometer. Rotations of the whole foot about three orthogonal axes relative to the shank were estimated using a five camera motion analysis system.

Findings. Biplanar insoles significantly ($P < 0.05$) reduced maximum eversion by an average of 3.1° when compared to the no insole condition. The cobra insole reduced maximum eversion by an average of 2.1° when compared to the no insole condition. This difference approached statistical significance ($P = 0.058$).

Interpretation. Biplanar and cobra insoles had no significant effect on maximum dorsiflexion, abduction or rate of eversion, when compared to the no insole condition. These results provide some limited support for the use of simple insoles to control for excessive foot pronation during walking.

Keywords: Gait analysis; Orthotic; Pronation

1. Introduction

Various studies have investigated abnormal pronation which is a combined movement made up of three planar movements namely, eversion, dorsiflexion and abduction. These studies have further indicated that this is a major contributing aetiological factor to lower limb musculoskeletal pathologies (Landorf and Keenen, 1999; Stacoff et al., 2000). Repetitive activities, which include excessive pronation of the subtalar and tarsal joints during midstance phase of gait, may lead to an increase in the load applied to the foot's ligaments, muscles and tendons, often resulting in chronic injuries. This pathological pronation of the foot has been suggested to increase joint range of motion and reduces the propulsive leverage needed to complete the gait cycle efficiently (Dennis et al., 1985; McCulloh et al., 1993).

Moulded foot orthotics have been shown to be a successful means of injury treatment, aiding in the dramatic reduction of patients' symptoms (Landorf and Keenen, 1999; McCourt, 1992). Donatelli et al. (1988) surveyed 53 patients after orthotic treatment and found that 96% reported pain relief and 52% of the patients would

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not leave the house without the devices in their shoes. Furthermore, Mundermann et al. (2003) highlight that the comfort of orthosis is a clear factor in the success of a device. Cavangh and Christopher (1989) reported that from a group of 96 injured runners, 64 gained relief of symptoms after receiving moulded orthotics.

Three-dimensional motion analysis has allowed the foot to be assessed in each plane of motion. Research using three-dimensional analysis has shown that moulded orthotics affect the rearfoot by significantly reducing maximum pronation angle and the total period of pronation (McCulloh et al., 1993; Nawoczenski et al., 1995; Stacoff et al., 2000). Nigg et al. (1998) found that medially posted moulded orthotics had a negligible effect on rearfoot motion during running. However, they had a significant effect on internal rotation of the tibia, reducing the total range of motion by a mean of 31%. This effect of orthotics on tibial rotation has been reported in other research studies (Nawoczenski et al., 1995; Nigg et al., 1999; Stacoff et al., 2000) suggesting that orthotics can have significant effects at other lower limb joints proximal to the foot.

Moulded orthotics are expensive and time consuming to produce as the construction of the device involves several manufacturing stages to provide the prescription. They can also be difficult to fit into footwear due to the bulk of the material (Chapman, 1999; McCourt, 1990; Stell and Buckley, 1998). A simple insole, flat non moulded device, (Fig. 1) is an alternative therapy to moulded orthotics. They are often chosen by practitioners to provide patients with immediate treatment; the simple insole being manufactured at the patient's initial consultation. Additionally, simple insoles can be used to provide treatment in footwear that has restricted room, such as specialised sports shoes. It was estimated by McCourt (1990) that simple insoles were 2.5 times cheaper than moulded orthotics to produce and could be manufactured in a third of the time. This enables clinicians to manufacture the devices in the work place and utilise the insoles as a preliminary indicator of assessment accuracy (Chapman, 1999).

Biplanar Insole Cobra insole

Fig. 1. Insole designs: Texon base with EVA posting material.

There are a number of corrective simple insoles that can be prescribed. The base material is frequently made from texon/regen (Texon Intl., Middlesbrough, UK) fibreboard and Ethyl Vinyl Acetate (EVA) is used as a posting material. The aim of a corrective device is to reduce the effects of abnormal pronation by influencing foot function. Biplanar wedges are commonly referred to as medial wedges and theoretically tilt the calcaneus in the frontal and sagittal planes, to limit eversion and dorsiflexion, respectively. It has been suggested that the inclination of the applied wedge corresponds to the clinical assessment of the patient's subtalar joint axis (Kirby, 1989). Stell and Buckley (1998) used two-dimensional video analysis to compare this type of insole to bespoke moulded orthotics and found that the simple insole was as successful in controlling subtalar joint pronation as the orthotic, during treadmill running. However, the results of this study should be treated with some caution as the motion of the foot was masked by a training shoe.

Cobra pads are the closest relations to moulded orthotics as they incorporate an arch profile and heel cup. Clarke and Fredrick (1983) described the effects of the pad as supporting the arch and controlling the motion of the calcaneus, by incorporating a medial post and a dome in the arch profile. The highest point of this insole is positioned under the sustamtaculum tali, an anatomical landmark on the talus, which has been suggested to be the most effective anatomical position to limit subtalar joint movement if supported (Nigg et al., 1999). Cobra insoles were originally designed for the treatment of female patients wearing court shoes so that minimal material was added to the footwear.

There has only been minimal research to support the kinematic effects of simple corrective insoles on rearfoot motion, despite them being widely used for patient care. The purpose of this study therefore was to establish the effect of two specifically selected simple insoles on three-dimensional foot motion during normal walking.

2. Methods

2.1. Participants

Nine active males aged 19–35 years (mean 27 years), body mass 70–87 kg (mean 77.5 kg) volunteered to participate in this study and gave written informed consent prior to any testing. Ethical approval was obtained from the Manchester Metropolitan University ethics committee.

All participants had an inverted whole foot position when the subtalar joint was placed in neutral. This was assessed using a Dynastat weight bearing goniometer (Dynastat, Stafford, UK). Neutral was defined as the most congruent articulation of the talus on the calcaneus where neither talar head could be palpated clinically (Sell et al., 1994). No participants had received any previous podiatric orthotic therapy. Exclusion from the study occurred if the whole foot was not inverted in a subtalar joint neutral position, if the subtalar joint axis was positioned medially to the plantar weight bearing area of the foot, as described by Kirby (1989), or if the navicular drop was less than 10mm. (McPoil and Cornwall, 1999). Any physical or neurological lower limb abnormalities also excluded participation.

2.2. Insole design

Two simple insoles (i) biplanar and (ii) cobra were manufactured for each participant (Fig. 1). The insoles were made from a standard template sizing system in Texon; posting was made from EVA shore A60. A digital biometer was used to ensure that the posting was accurately manufactured to the correct angle measured from the participant's clinical assessment. Adjustable trekking sandals were used as footwear and insoles were securely fixed on the inside of the sandal using Velcro (Fig. 2).

2.3. Data collection

Each participant wore swimming trunks so that the leg was visible for marker allocation. Reflective markers were placed on a total of six landmarks on the left limb; three markers on the foot and three on the shank. The markers were used to define a right-handed local coordinate system for each segment. These were labelled the foot coordinate system (FCS) and the shank coordinate system (SCS), as shown in Fig. 3.

The movement of the limb markers were tracked using five infra red cameras (Motion Analysis Corporation, Santa Rosa, CA, USA) sampling at 120 Hz, during normal walking. Each camera was focused on a volume that had been calibrated statically and dynamically for three-dimensional analysis using a standardised 1 m^3 calibration frame and a 1 m wand. A force platform sampling at 480 Hz (AMTI Inc., Watertown, MA, USA) was incorporated into the data collection area and synchronised with the motion analysis system to accurately identify when heel strike occurred.

Fig. 2. Trekking sandal with adjustable velcro fastening.

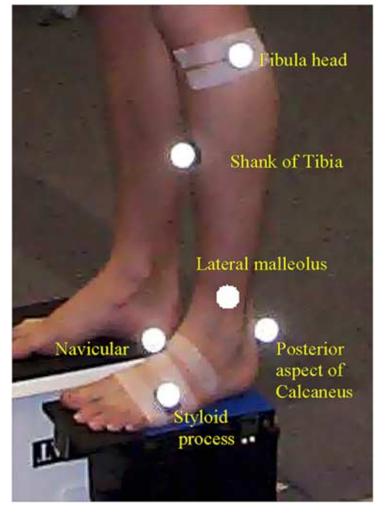

Fig. 3. Landmarks on the left limb to define a SCS and FCS whilst the foot is placed in neutral on the Dynastat.

A neutral trial, for each participant, was recorded by the five cameras to establish neutral joint angles (Fig. 3). This involved the participant standing motionless on the Dynastat goniometer in their measured subtalar joint neutral position, with their knee in full extension and lower limb vertical. Each participant was then given time to become familiarised with the footwear and was allowed a number of walking trials prior to data collection. Data were collected for three test conditions: sandal only, sandal and biplanar insole, sandal and cobra insole. Five trials, one foot contact on the force platform, were performed under each condition at the participant's normal walking speed. A valid trial consisted of the participant striking their heel on the force platform without visually altering their gait.

2.4. Data processing

The three-dimensional coordinates of the nine limb markers were smoothed using a Butterworth filter with a cut off frequency of 7 Hz. All joint angles during the gait cycle were calculated and expressed relative to the joint angles measured in the neutral trial. Joint coordinate system (JCS) angles were calculated for the foot and knee (Kintrak 6.0). All foot angles were expressed relative to the SCS. Inversion/Eversion of the foot was defined as rotation of the FCS about its long axis, plantar flexion/dorsiflexion of the foot was rotation of the FCS about its sagittal axis and adduction/abduction was rotation of the FCS about its transverse axis.

2.5. Definition of variables

The following four dependent variables were obtained from the JCS angles for each participant performing five walking trials in each of the three insole conditions:

- *Maximum dorsiflexion angle* (°)—the maximum dorsiflexion angle of the foot during ground contact.
- *Maximum eversion angle* (°)—the maximum eversion angle of the foot during ground contact.
- *Maximum abduction angle* (°)—the maximum abduction angle of the foot during ground contact.
- *Maximum eversion velocity* ($^\circ s^{-1}$)—the maximum rate of change of the eversion angle as a function of time.

2.6. Statistical analysis

All statistics were computed using SPSS version 10.0 for Windows. Four one-way ANOVAs for repeated measures were used to test for differences between the three insole conditions. The level for statistical significance was set at $P < 0.05$.

Examination of the variances for each of the dependent variables showed that F-max was always less than 3. Therefore, the assumption of homogeneity of variance was not violated. The assumption of Sphericity was assessed using Mauchly's test. The value for Mauchly was 0.595 and was not significant ($P = 0.163$). Where a significant difference was found, the Bonferroni post hoc test was used to identify where the difference lay.

3. Results

This section presents the mean data from the nine participants obtained for the three insole conditions. Each value reported therefore represents the mean of 45 measurements. Fig. 4 illustrates the maximum eversion angles exhibited by the participants under the three conditions.

All nine participants exhibited less eversion when walking with biplanar insoles, compared to walking without insoles. Biplanar insoles reduced maximum eversion by an average of 3.1° (range 0.1–5.0°) when compared to the no insole condition. Post hoc analysis revealed that this reduction was statistically significant ($P < 0.05$), and a moderate effects size of 0.62 was obtained. Eight of the nine participants displayed less eversion when walking with cobra insoles, compared to walking without insoles. The cobra insole reduced maximum eversion by an average of 2.1° when compared to the no insole condition. Although this result was not statistically significant ($P = 0.058$), only a small to moderate Effects Size of 0.46 was found. No significant difference was found between the biplanar and cobra insole conditions.

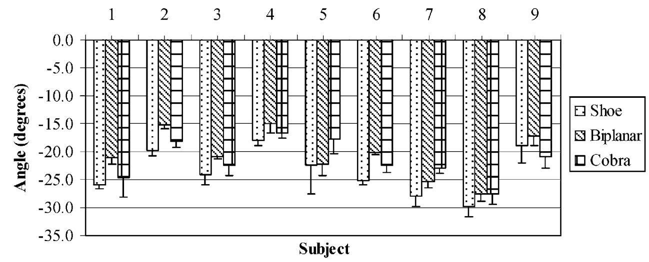

Fig. 4. Maximum eversion angle for the left foot of each subject during walking under the three insole conditions.

The biplanar and cobra insoles had no significant effect on maximum dorsiflexion or maximum abduction, as illustrated by Figs. 5 and 6, respectively. There was less than a 1.0° difference between the mean values for the three insole conditions, for both of these variables. No single participant exhibited more than a 2° difference in his maximum dorsiflexion, or a 1.5° difference in maximum abduction, across the three insole conditions.

When walking without an insole, the participants had a maximum eversion velocity of 95 $^\circ s^{-1}$ (SD 12$^\circ s^{-1}$) This eversion velocity did not change significantly when

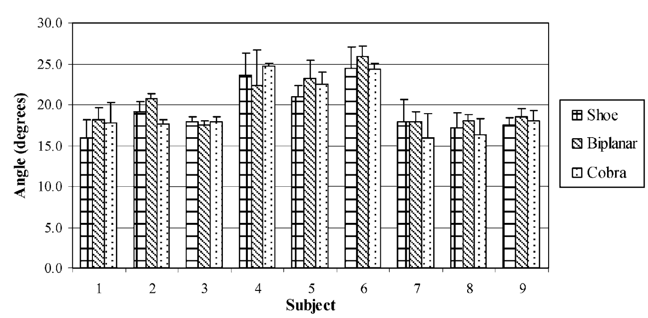

Fig. 5. Maximum dorsi flexion angle for the left foot of each subject during walking under the three insole conditions.

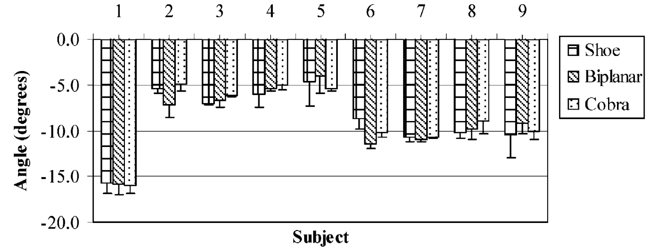

Fig. 6. Maximum abduction angle for the left foot of each subject during walking under the three insole conditions.

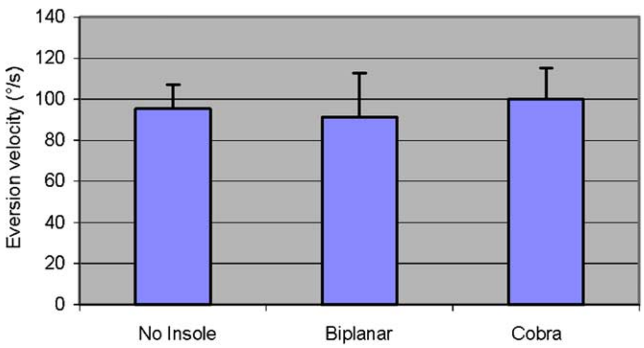

Fig. 7. Maximum eversion velocity (mean ± standard deviation) for the left foot during walking under the three insole conditions.

they used either a biplanar (91 (SD 22)$^\circ s^{-1}$) or a cobra (100 (SD 15)$^\circ s^{-1}$) insole (Fig. 7).

4. Discussion

The results indicate a significant reduction in eversion whilst using a biplanar simple insole, but no significance from using a cobra equivalent insole. Since, inversion/eversion of the foot has been highlighted as the greatest component of supination/pronation (McCulloh et al., 1993), a reduction in eversion could be directly linked to a reduction in pronation. (Blake and Ferguson, 1993; Landorf and Keenen, 1999; Stacoff et al., 2000).

The results of this study conform with previous studies on maximal eversion angle (McCulloh et al., 1993; Nigg et al., 1998) which have indicated a reduction of 3–4° in maximum eversion with the use of medial wedges. Although Nigg et al. (1998) questioned the relevance of a 3–4° change in eversion angle, the results of this study show that the eversion angle is the only significant change made by medial-wedged insoles. Patient's symptoms have been shown to improve with the use of such devices (McCourt, 1990; McCourt, 1992) and although it is clear that there are other major factors that influence the success of insole treatment (Mundermann et al., 2003) this small angle change observed in this study could be a contributing factor.

The failure of the dorsiflexion angle results to differentiate between conditions suggests that dorsiflexion angle is not a substantial component in pronation that is influenced by antipronatory devices. However, dorsiflexion of the foot occurs mainly at the ankle joint, although motion can also be seen at the subtalar and other lesser tarsal joints (Engsberg and Andrews, 1986). Marker placements used within this study did not allow for any specific joint in the foot to be identified, as the method adopted isolated the foot as a single rigid segment model. Therefore any changes in dorsiflexion at smaller joints in the foot were masked by gross motion of the ankle joint. A more accurate assessment of the subtalar joint would be difficult with this method of analysis as the key points to be marked would be too close to each other and it would be troublesome in identifying the markers within this particular analysis program.

In absolute terms the results seen for dorsiflexion differ substantially from previous studies in this area. Although, McCulloh et al. (1993) measures a typical dorsiflexion angle of 11° (SD 1.9°), which is substantially lower than the values recorded within this study, this effect could possibly be caused by gaining a neutral subtalar joint position for each subject on the Dynastat. By using this benchmark for a zero reference point the ranges of motion in dorsiflexion will be higher than other reports as each subject's foot assessed in subtalar joint neutral will be neither dorsiflexed or plantarflexed. Whilst such a variation in angles does not invalidate the results of this study it is recommended that these possible causes of the variation are investigated further.

None of the previous studies in this area have independently measured abduction angle in their research. Whilst this angle was measured in this study, the similarity of results between conditions suggests that abduction angle is not influenced by antipronatory treatments. This conforms with existing understanding of the pronation components and their relative ratios, and therefore suggests that abduction angle is not a meaningful determinant of subtalar joint pronation (McCulloh et al., 1993).

Eversion velocity was reduced when using a biplanar insole however this reduction was not deemed significant. Past research has debated the effect of reducing eversion velocity with orthoses and the significance of such is still unclear (Landorf and Keenen, 1999).

The widely accepted theory that the biplanar insole reduces motion in two planes however was not supported by this research. Although eversion was reduced the insole did not alter dorsiflexion. This could be due to the design of data collection as rotations of the foot were based on a single hinged model.

The theory behind the cobra insole described by Clarke and Fredrick (1983) is to be questioned with the results from this study. There was no significant alteration in eversion, dorsiflexion or abduction, therefore highlighting that the cobra does not significantly control the calcaneus as previously suggested. If there was abnormal movement of the calcaneus on the curved heel cup of the insole this could account for insignificant changes. Although the sandals used in this research had strong ankle straps there was minimal heel support. Repeating the trials with footwear that had a heel cup may have altered these results by reducing the slipping of the foot on the device.

The primary limitation of this study is the small sample size. In addition, a direct comparison of the results with previous studies is further complicated due to the different analytical methods and terminology used. A universal set of terms would enhance research in this area as direct comparisons of angles and velocities can be significantly compared.

5. Conclusion

Biplanar insoles have been shown to significantly reduce the eversion angle of the foot. Moulded orthotics are most frequently researched for cause and effect but not always used in clinical practice. Other treatment modalities including simple insoles are clinically used as an alternative to moulded orthotics. The results here highlight that biplanar simple insoles do reduce foot motion adding to the continued research that goes into why orthotics and insoles help to reduce patients symptoms.

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