Influence and benefits of foot orthoses on kinematics, kinetics and muscle activation during step descent task

Abstract

Medial wedged foot orthoses are frequently prescribed to reduce retropatellar stress in patients with patellofemoral pain (PFP) by controlling calcaneal eversion and internal rotation of the tibia. During activities of daily living, the highest patella loads occur during stair descent, but the effect of foot orthoses during stair descent remains unclear. The purpose of this study was to compare the kinematics, kinetics and muscle activation during a step descent task in healthy volunteers using three designs of foot orthoses (insoles). Sixteen healthy subjects with a mean age of 25.7 years, BMI of 23.3, and +5 Foot Posture Index were recruited. Subjects performed a step down task from 20 cm using a 5^{o} rearfoot medial wedge (R), a 5^{o} rearfoot and forefoot medial wedge (R/F), and a control flat insole (C). Significant improvements in control were seen in the R and R/F insoles over the C insole in the foot and at the ankle and hip kinematics. The R and R/F insoles increased the knee adduction moments, but decreased the knee internal rotation moment compared to the C insole. Adductor hallucis (AH) activity was reduced with both insoles, whereas tibialis anterior (TA) activity was reduced with the R insole only. Foot orthoses can change joint mechanics in the foot and lower limbs providing greater stability and less work done by AH and TA muscles. This data supports the use of foot orthoses to provide functional benefits during step descent, which may benefit patients with PFP.

1. Introduction

The human musculoskeletal system is challenged daily across different types and levels of terrain [1]. Stairs are commonly encountered in the workplace, at home, and in the community. Although these are rarely challenging for healthy individuals, they could be considered a difficult activity of daily living for elderly, injured or disabled persons where motor function is compromised [2].

Many studies have demonstrated significant differences between stair climbing and level walking. In particular, step descent has been shown to produce greater moments and range of motion at the knee leading to a significantly greater mechanical demand [2]. Compared to step ascent, a step descent is more challenging due to the center of mass being moved both forwards and down in a controlled lowering phase [3]. This is achieved through eccentric muscular activation, which controls the rate of lowering the center of mass [3]. In addition, during the controlled lowering phase, the knee joint starts from a relatively stable extended position and flexes towards an increasingly unstable position. The increased joint flexion causes a progressive increase in the external flexion moment which is matched by progressively increasing eccentric muscle activity and joint forces in order to prevent collapse [3]. In healthy adults, stair descent has been shown to yield greater forces and greater peak knee abduction moments compared to stair ascent and level walking [4,5]. Differences in the foot-to-ground interface have also been observed, with a heel-to-toe contact pattern during level gait and a toe-to-heel contact during stair climbing. Specifically, the metatarsal heads accept the load followed by a lowering onto the heel [6]. An additional difference between level walking and descending stairs is linked to the vertical ground reaction force (vGRF). Both tasks show two peaks in vGRF; however, during the step descent the first vertical peak is greater than the second peak [1], and although the horizontal forces are similar for braking impulse, the propulsive force is lower during step descent [1].

Changes in movement control in the feet and lower limbs may lead to compensations during functional activities and could directly be associated with risk factors and the occurrence of injuries [7]. According to Borque et al. [8], activities that include an inclined and irregular surface such as stair descent, can demonstrate symptoms which may arise from the need to compensate for inherent instabilities in the musculoskeletal system.

Foot motions such as excessive supination and pronation are transferred proximally up the lower limb, in particular through the rearfoot torque mechanism. This can generate an increased demand on structures such as the anterior cruciate ligament and patellofemoral joint [7,9]. The use of insoles to correct foot alignment is a common conservative management approach that aims to act as a mechanical barrier against excessive patterns of movement of the foot. Medial wedged foot orthoses are often prescribed to reduce the knee and hip joint loads thought to increase retropatellar stress by reducing calcaneal eversion and tibial internal rotation [9]. Clinically, medial wedges are frequently positioned under the rearfoot acting together with the arch support. However, recent studies have suggested that both the rearfoot and forefoot influence the control of movements in excessive pronation. For example, Resende et al (2015) noted that an increase in forefoot pronation may result in increased rearfoot eversion internal hip rotation during gait [10]. In an earlier study, Monaghan et al. (2014) showed that the angle of the forefoot to the ground at forefoot contact determined the amount and duration of eversion during walking. It was noted that the rearfoot angle at rearfoot contact had no effect on the amplitude or duration of eversion during walking [11]. In addition, Rodrigues et al. (2013) demonstrated that medial wedge insoles in the forefoot and rearfoot reduced eversion and eversion velocity of the ankle joint complex in runners, with and without anterior knee pain [12]. Due to the toe-to-heel contact pattern of movement during the step descent task, postings of orthoses under the forefoot could play an important role in the control of foot pronation and associated movements of proximal joints of the lower limb.

The purpose of this study was to compare the kinematics, kinetics and muscle activation during the lowering phase of the step descent task of healthy volunteers using three designs of insoles (foot orthoses). These included; a 5^{o} medial wedge insole positioned under the rearfoot, a 5^{o} medial wedge insole positioned under the rearfoot and forefoot, and a control flat insole.

2. Methods

2.1 Participants

Sixteen healthy subjects (10 males and 6 females) with a mean age of 25.7 years (SD 5.8), body weight of 71.7 kg (SD 10.6), height of 174.8 cm (SD 9.2), mean BMI of 23.3 (SD 1.7) and a mean score of +5 (SD 4) for the Foot Posture Index (FPI) (version 6) were recruited. All participants were free of previous and present history of patellofemoral pain, injuries to the lower-limbs or pelvis or surgery. The volunteers signed an informed consent in accordance with the Declaration of Helsinki. This study was approved by the Ethical Committee of the University of Central Lancashire.

2.2 Procedures

An initial assessment was conducted which included measures of body weight, height and Foot Posture Index (version 6) [1]. Lower limb kinematic data were then obtained using a 10-camera Oqus 7 system at 100 Hz (Qualisys Medical AB, Gothenburg, Sweden). Passive retro-reflective markers were placed on the lower limbs and pelvis using the Calibrated Anatomical System Technique allowing the segmental kinematics to be tracked in 6-degrees of freedom [13]. Anatomical markers were positioned by the same researcher on the anterior superior iliac spine, posterior superior iliac spine, greater trochanter, medial and lateral femoral epicondyle, medial and lateral malleoli and over medial and lateral aspects of 1st and 5th metatarsal respectively. Additionally clusters of non-collinear markers were attached to the shank and thigh, and markers were also placed over rearfoot, midfoot and forefoot aspects of the shoes [13]. Static calibration trials were obtained with the participant in the anatomical position. Kinetic data were collected using two AMTI force plates at 2000 Hz (Advanced Mechanical Technology Inc, Watertown, MA). Joint moment data were calculated using three-dimensional inverse dynamics, and the external joint moment data were normalised to body mass (N m/kg).

In addition, electromyographic (EMG) data were obtained from the tibialis anterior (TA), peroneus longus (PL), medial gastrocnemius (MG) and abductor hallucis muscles (AH) using a Trigno Wireless EMG system at 2000 Hz (Delsys Inc., Boston, MA). The skin was cleaned with alcohol wipes and the standard EMG electrodes were positioned in accordance with the SENIAM guidelines and fixed with double-sided adhesive skin interfaces. Standard Trigno wireless sensors were used to collect data from TA, PL and MG muscles, and a Trigno Mini wireless sensor was placed such that the data from AH could be collected inside the footwear with minimal sensory disturbance (Figure 1). The AH was palpated, and the mini electrodes were fixed with a double-sided adhesive skin interface and the connecting wire was secured with Hyperfix.

(Insert Figure 1 here)

Data were collected from the dominant lower limb and pelvis, with the dominant limb being defined as the limb with which they would kick a ball. Six repetitions of a 20 step descent at a self-selected speed were performed from a step positioned on the first force plate, which has been previously been used to assess closed chain eccentric control and stability [3], were performed under three randomized conditions. The conditions included; control flat insole (C-insoles), a medial longitudinal arch support with a 5 degree medial rearfoot posting insole (R insoles), and a medial longitudinal arch support with a 5 degree medial forefoot and rearfoot posting (R/F insoles). The base of the insoles was pre-fabricated with a standardised arch support and neutral heel. The 5 degree posting material was made from ethylene vinyl acetate (EVA) and was affixed under insoles using double side tape (Figure 2). All volunteers wore appropriately sized standardized footwear (Dr Comfort Winner Plus). The size of the insoles was adjusted to fit the footwear, however the insoles were not customized for each volunteer.

(Insert Figure 2 here)

2.4 Data processing

Raw kinematic, kinetic and EMG data were exported to Visual3D (C-Motion Inc., Germantown, USA). Kinematic and kinetic data were filtered using fourth order Butterworth filters with cut off frequencies of 6Hz and 25Hz, respectively. EMG data were zeroed, band-pass filtered with corner frequencies of 20Hz and 500 Hz, full-wave rectified and enveloped using a fourth-order low-pass Butterworth filter with a cut-off frequency of 25Hz. The EMG data were normalized to the maximal observed signal during the dynamic contraction during the movement tasks [6]. For all data, time was normalized to 101 points from toe-off of the non-dominant foot, using the threshold of the 1st metatarsal marker vertical trajectory velocity, to initial contact of the contralateral limb using a threshold of 10N on vertical force on the second force plate.

The mean kinematic, kinetic and EMG values for each condition were recorded during the single limb descent phase. The dependent kinematic variables included; the minimum, maximum and range of motion in the sagittal, coronal and transverse plane at the forefoot, midfoot, ankle, knee and hip. The dependent kinetic variables included peak ankle, knee and hip moments in all three planes. Finally, the dependent variables for the peak and integrated EMG (iEMG) values from TA, PL, MG and AH were found. These were then normalized to the maximal observed signal during single limb descent phase for each muscle [14]

2.5 Data Analysis

Each kinematic and kinetic variable was assessed and found to be normally distributed and suitable for parametric statistical testing. Repeated measure ANOVAs with pairwise comparisons were performed to compare the three insole conditions, in addition the effect size ($p\eta^2$) was also found. Bonferroni corrections were employed to allow for multiple comparisons and to reduce the possibility of type I errors. All statistical calculations were conducted using SPSS v.22.0 (SPSS Inc., Chicago, USA), with the α level set at 0.05.

3. Results

Descriptive statistics are presented in Table 1 and time series curves for each variable with a statistically significant difference are presented in Figure 3. The Repeated Measured ANOVAs showed significant differences between the conditions during the controlled lowering phase of the step descent task, for kinematics at the foot, ankle and hip (Table 1). Further pairwise comparisons showed significant differences between conditions for the foot, ankle and hip (Table 2). At the foot and ankle, significantly less metatarsocalcaneal internal rotation ($p=0.002$, $p=0.030$), and calcaneal eversion ($p<0.001$, $p=0.003$; $p=0.006$, $p=0.014$), were seen with the R and R/F insoles compared to the C. In addition, ankle abduction was also significantly reduced using the R and R/F insoles compared to the C insole ($p=0.009$, $p=0.001$; $p=0.018$, $p=0.007$). At the hip, initiation of motion occurred with significantly less peak hip external rotation for the C insoles ($p=0.007$, $p=0.002$). All insoles showed statistically different results for hip internal rotation ($p=0.018$, $p=0.003$, $P=0.023$), with the C insoles showing the greatest rotation. In the coronal plane the R insole produced a significant lower hip adduction ($p<0.001$), and lower hip coronal plane range of motion compared to C and R/F insoles ($p=0.001$, $p=0.008$).

The repeated measures ANOVAs also showed differences in the knee moments and iEMG activity for the TA and AH between the conditions (Table 1). Further pairwise comparisons showed significantly lower knee adduction moment for the C insoles compared to the R/F and

R insoles of the step descent task ($p<0.001$). However, the C insoles showed a greater knee internal rotation moment in relation to the R/F and R insoles ($p<0.001$). The iEMG data showed a higher peak of activity and iEMG in AH with the C insoles compared to the R/F and R insoles ($p=0.001$, $p=0.013$; $p=0.015$, $p=0.006$), and lower TA iEMG with the R insoles compared to the R/F and C insoles ($p=0.001$, $p=0.01$) (Table 3).

4. Discussion

This study compared the kinematics, kinetics and muscle activation during the lowering phase of the step descent task of healthy volunteers using three designs of insoles. The results showed a significant reduction in over pronation of the foot and associated coupled movements on lower limbs wearing the R and R/F insole compared to the C insole.

During closed chain activities, joint motions of the lower limb are interdependent, and excessive movements from one joint may overload tissues in the kinematic chain [15]. Studies have focused on the clinical relevance of excessive pronation during walking, which has been identified as a major factor in the development of overuse injuries and PFP [12,16]. This is usually associated with a lack of muscle strength, stability, and overuse of the foot muscles due to the oblique angle of the subtalar joint [10,17]. Insoles (foot orthoses) are a common modality employed by many clinicians with evidence indicating that they can prevent overuse conditions of lower limbs [18]. Whilst the mechanism of action is debated, Hamlyn et al. previously noted that insoles with MLA support increased the area of contact under the feet reducing excessive pronation in individuals with functional ankle instability [19]. Although, in this current study the foot posture (FPI) was assessed and found to be +5 with a standard deviation of 4, indicating that on average the sample did not present with excessive pronation [20]. However, the results did show significant reductions in metatarsal to calcaneal internal rotation, ankle eversion and ankle abduction in the R and R/F compared to the C insole. This indicates that the insoles with the medial longitudinal arch (MLA) support offered greater control of the movements commonly associated with excessive pronation, and reduced work done by the AH and TA muscles. These muscles have been shown to play a role in stabilizing the MLA and consequently the pronation of the foot, which has been linked to an increase in load and MLA compression [21]. During walking, the TA's main action is eccentric control of the foot during weight acceptance, and potentially influencing the rate of rearfoot eversion [22,23]. Theoretically, forefoot motion will also be influenced by the TA, as its actions are described as elevation and lateral rotation of the 1st metatarsal and medial cuneiform, as well as raising the MLA during push-off [22]. Although Cornwall et al. considered the TA to be active from before the foot strikes the ground until the foot is flat [23], during a step descent the TA activity, lower limb kinematics and kinetics data show a consistently higher coefficient of variation compared to walking gait [24].

iEMG of the AH muscle has been studied by Reeser et al. who demonstrated significant myoelectric activity during late stance and the toe-off phase of gait [25]. To date, no studies have analyzed AH muscle activity during step descent. This study showed significantly lower TA and AH activity with greater joint control when using the R and R/F insoles compared to the C insole during the step descent. This may indicate better foot position control during step descent and improved support by the insoles and requires further study.

Pierrynowski showed that female participants with PFP descended with the hip more adducted and internally rotated compared to asymptomatic individuals during a step-down activity [26]. Evidence indicates that a relationship exists between excessive eversion of the rearfoot and PFP [27,28]. From a theoretical perspective, excessive eversion of the rearfoot causes excessive internal rotation of the tibia, which consequently creates a higher internal rotation and adduction of the hip. Adduction and internal rotation of the hip are reported to increase dynamic knee valgus and lateral tracking of the patella, which can lead to a reduction in the contact area of the patella, which increases PFP and stress [15]. This study observed a reduction in internal rotation and hip adduction with R and R/F insoles compared to the C insole. The R insole showed a smaller reduction compared to the R/F insole and a lower hip range of motion (coronal plane), however it appears that R and R/F insoles can influence the control of excessive movement being transferred proximally.

The knee joint plays an important role in controlling the movement and during step descent [13,29] with the hip and ankle indirectly influencing knee kinematics [10]. Previous studies have shown that dynamic knee valgus can predispose lower limb injuries, especially at the knee, where step descent activity may be an important clinical outcome. The presence of alterations in movement patterns, distally and proximally, can generate overload in the lower limb joints contributing to musculoskeletal pain [30]. In this current study, R and R/F insoles significantly decreased internal rotation and knee adduction compared to the C insole which has the potential to benefit individuals who have PFP. Whilst the findings of this study showed that F and R/F insoles can induce significant functional improvements, further evaluation is need on individuals with PFP pain and in individuals with different foot postures. In addition, this current study has reported on a 20 cm stepdown task, rather than continuous stair descent, which may yield different results [24].

Foot orthoses can change joint mechanics in the foot and lower limbs providing greater stability and less work done by AH and TA muscles. This data supports the use of foot orthoses to provide functional benefits during step descent, which may benefit patients with PFP. It should be noted however, that these results were obtained from healthy volunteer participants and this work must be replicated in patients diagnosed with PFP.

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