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

Douglas Bonifácio[a], Jim Richards[b], James Selfe[c], Sarah Curran[d], Renato Trede[a,*]

[a] Postgraduate Program in Rehabilitation and Functional Performance, UFVJM, Diamantina, Brazil
[b] Allied Health Research Unit, University of Central Lancashire, Preston, UK
[c] Department of Health Professions, Faculty of Health, Psychology and Social Care, Manchester Metropolitan University, Manchester, UK
[d] Cardiff School of Sport and Health Sciences, Cardiff Metropolitan University, UK

A R T I C L E I N F O

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A B S T R A C T

Background: 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.
Research question: 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).
Methods: 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 5o rearfoot medial wedge (R), a 5o rearfoot and forefoot medial wedge (R/F), and a control flat insole (C).
Results: 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 reduced knee internal rotation moment compared to the C insole. Abductor hallucis (AH) activity was reduced with both insoles, whereas tibialis anterior (TA) activity was reduced with the R insole only.
Significance: 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 of 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 yield greater forces and greater peak knee abduction moments compared to stair ascent increasing eccentric muscle activity and joint forces in order to prevent collapse [3]. In healthy adults, stair descent has been shown to 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

* Corresponding author at: Postgraduate Program in Rehabilitation and Functional Performance, UFVJM, Diamantina, Brazil.
E-mail address: renato.trede@gmail.com (R. Trede).

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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. 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. 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. 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° medial wedge insole positioned under the rearfoot, a 5° 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). 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 (Nm/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 (Fig. 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.

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 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° medial rearfoot posting insole (R insoles), and a medial longitudinal arch support with a 5° medial forefoot and rearfoot posting (R/F insoles). The base of the insoles was prefabricated with a standardised arch support and neutral heel. The 5° posting material was made from ethylene vinyl acetate (EVA) and was affixed under insoles using double side tape (Fig. 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.

2.3. Data processing

Raw kinematic, kinetic and EMG data were exported to Visual3D (C-

Fig. 1. Trigno Mini wireless sensor placed on the foot for abductor hallucis muscle.