The effect of anti-pronation foot orthoses on hip and knee kinematics and muscle activity during a functional step-up task in healthy individuals: A laboratory study

Simon Lack, Christian Barton, Peter Malliaras, Richard Twycross-Lewis, Roger Woledge, Dylan Morrissey

Centre for Sports and Exercise Medicine, William Harvey Research Institute, Bart’s and the London School of Medicine and Dentistry, Queen Mary University of London, United Kingdom

A R T I C L E I N F O

Article history:
Received 25 March 2013
Accepted 20 November 2013

Keywords:
Orthoses
Onset times
Kinematics
Electromyography
Patellofemoral pain
Anti-pronation foot orthoses

A B S T R A C T

Background: Greater frontal and transverse plane hip and knee motion, and delayed gluteus medius and vastus medialis oblique activation have frequently been identified in patellofemoral pain syndrome populations, whilst prefabricated anti-pronation foot orthoses have been reported to reduce symptoms. The aim of the study was to evaluate the effects of such orthoses on hip and knee kinematics, gluteal and vasti muscle activity, kinematic and electromyographic interactions alongside correlations with specific clinical measures.

Methods: Eighteen asymptomatic individuals (11 male 7 female) had measures taken of static foot posture and ankle range of motion. Hip muscle activity and kinematics were measured using electromyography and an active motion capture system during a step-up task. Order of testing with or without orthoses was determined using a coin toss.

Findings: Between condition paired t-tests indicated significantly reduced peak hip adduction angles (1.56°, \( P = 0.013 \)) and significantly reduced knee internal rotation (1.3°, \( P < 0.05 \)) in the orthoses condition. Reduced ankle dorsiflexion range of motion correlated with a reduction in hip adduction following the orthoses intervention (\( r = 0.59, P = 0.013 \)).

Interpretation: The effects of prefabricated orthoses may be partially explained by kinematic alterations that occur proximal to the foot in the kinetic chain. These clinically and biomechanically relevant effects appear more evident in those with reduced underlying ankle motion. Further research is indicated using a symptomatic population to explore the clinical relevance of these observations.

© 2013 Published by Elsevier Ltd.

1. Background

Patellofemoral pain (PFP) is a prevalent complaint within physically active populations (Baquie and Brukner, 1997; Taunton et al., 2002) and is reported to be one of the most common injuries of the lower extremity (Baquie and Brukner, 1997; Taunton et al., 2002). Among 2002 patients presenting to a sports medicine clinic with running related injuries, 842 (42.1%) reported knee pain with 331 (46%) being diagnosed with PFP (Taunton et al., 2002). Although the anatomical source of pain is uncertain (Powers et al., 2012), the aetiology of PFP is considered to be multifactorial, with numerous risk factors identified (Heino Brechter and Powers, 2002; Powers et al., 2012).

Consensus among clinicians and researchers is that PFP can develop as a result of altered or elevated lateral patellofemoral joint (PFJ) loading with distal, proximal and local biomechanical factors thought to contribute (Powers et al., 2012). Distally, it has been proposed that excessive sub-talar joint pronation (Tiberio, 1987) might result in greater tibial segment and hip joint internal rotation (Fig. 2). External tibio-femoral joint rotation, necessary to extend the knee and align the foot in the direction of travel, consequently results in increased lateral loading of the PFJ. This proposed kinematic coupling between lower limb segments was recently supported by reports that greater peak rearfoot eversion was associated with greater tibial internal rotation in individuals with PFP (Barton et al., 2012). Additionally, imposed constraint of ankle dorsiflexion range has been reported to result in decreased knee flexion angles during squatting activities (Macrum et al., 2012) identified prospectively as a risk factor for developing PFP (Boiling et al., 2009).

A growing body of research has explored neuromuscular and biomechanical variables at the hip in individuals with PFP (Aminaka et al., 2011; Nakagawa et al., 2012; Willson et al., 2011), reporting evidence of delayed GMed onset (Aminaka et al., 2011; Nakagawa et al., 2012) and increased hip adduction angles (Willson et al., 2011) during functional tasks. With favourable outcomes reported following a proximal strengthening intervention in PFP (Fukuda et al., 2012), further research exploring the effects at the hip of other commonly used interventions for managing PFP is clearly warranted.

Locally, neuromotor patterns of the quadriceps muscles have been proposed to contribute to altered lateral PFJ loading through the delay of vastus medialis oblique (VMO) compared to vastus lateralis (VL) (Fig. 1) (Coqueiro et al., 2005). A recent systematic review identified a...
trend of delayed VMO activation in individuals with PFP, although a substantial degree of heterogeneity across pooled studies and participants within studies was identified (Chester et al., 2008).

Physiotherapy intervention including patellofemoral joint (PFJ) mobilisation, patella taping, quadriceps strengthening and education remains the gold standard management of PFP management (Collins et al., 2008; Crossley et al., 2002). However, Level 1 evidence also exists to support the use of anti-pronation foot orthoses (APFOS) as an adjunct to hasten recovery (Collins et al., 2008). Specifically, Collins et al. (2008) randomised controlled trial reported significantly greater global improvement on a five point Likert scale at six weeks in a group of PFP individuals receiving prefabricated APFOS compared to a control group. However, the mechanism for this effectiveness is poorly understood, due to a paucity of research evaluating biomechanical effects of APFOS in individuals with PFP (Mills et al., 2010). Eng and Pierrynowski’s (Eng and Pierrynowski, 1994) study remains the only published paper exploring the effects of APFOS on lower limb kinematics, demonstrating a significant reduction in both frontal and transverse plane movements at the foot and knee. However, no attempt was made to evaluate effects proximal to the knee.

Limited literature evaluating the effects of APFOS on muscle activation at the hip also exists. Hertel et al. (Hertel et al., 2005) studied healthy adults during a single-leg squat and lateral step down tasks in neutral, medial and lateral posted orthoses and reported significantly increased EMG amplitude within both the VMO and gluteus medius (GMed) muscles for all conditions compared with no orthoses. However, without kinematic data, conclusions about the relevance of this to the control of lower limb alignment cannot be made.

The aim of this study was to improve understanding of the effects of APFOS on relevant hip and knee muscle activity and kinematics in normal subjects, as a precursor to identify the possible role of APFOS in the management of PFP populations. The hypotheses of this study were that: (i) APFOS would result in earlier onset of GMed and VMO, and a relatively delayed onset of VL, and a reduction in the peak frontal and transverse plane kinematics at the hip and knee during a functional step up task, (ii) foot posture index (FPI) scores indicating greater pronation and reduced ankle dorsiflexion range would influence kinematic coupling within the lower limb such that correlations with these biomechanical changes will be observed.

2. Method

2.1. Participants

A convenience sample of eighteen physically active, asymptomatic individuals (M 11 F 7; Mean (SD); age = 29.2 (3.7) years; height = 174.8 (7.2) cm; weight = 72.5 (11.8) kg) was recruited to participate in the study in response to advertisements within the university campi. Ethical approval was obtained from the Queen Mary University Ethics Committee and each participant provided written informed consent. Participants were required to have no history of lower extremity injury or knee pain in the last 12 months, and to be free of lower back pain or other neuro-musculoskeletal deficits potentially affecting stair ascent ability.

3. Procedure

3.1. Foot posture index

Static foot posture was assessed by the lead author, having previously tested over 30 individuals clinically, and deemed proficient using the six
Participants were randomised into either APFOS performing up to command. The dominant leg was always the lead leg, with all participants on a wooden step (combined height 22 cm), in response to a verbal force plate (Type 9281CA, Kistler Corporation, Switzerland) mounted asymptomatic controls having been successfully identified using four CodaMotion Cx1 sensor units (Charnwood Dynamics, Natwick, MA, USA) for post-processing.

3.2. Ankle dorsiflexion range

Knee bent ankle dorsiflexion (KBAD) range was achieved using digital inclinometer methods previously described (Munteanu et al., 2009). The dominant foot was placed along a taped line on the floor perpendicular to the wall. Participants lunged as far forward as possible whilst keeping the dominant heel on the ground and the knee bent. The largest angle of three measures was recorded from the inclinometer (Baseline® Digital Inclinometer 12-1057) placed on the anterior tibia.

3.3. Electromyographic recordings

Muscle activity of VMO, VL and GMed of the participant’s dominant leg was recorded by wireless surface electromyography (sEMG) (Telemyo 2400TG2, Noraxon, USA). The subject’s skin was prepared and pairs of Ag/AgCl surface electrodes with an intra-electrode distance of 20 mm (Tyco Healthcare, Germany) were placed over the muscles of interest according to standard SENIAM guidelines (Freriks et al., 2000). The GMed electrode was placed halfway along the line between iliac crest and greater trochanter, orientated vertically. The VMO electrode was placed at 80% of the distance down the line between the anterior superior iliac spine (ASIS) and the medial joint line just anterior to the medial ligament, orientated ~55° to the vertical. The VL electrode was placed 2/3 of the distance down the line from the ASIS to the lateral patellar border, orientated ~15° to the vertical. Surface EMG signals were sampled at 1500 Hz, pre-amplified and band-pass filtered between 10 and 500 Hz, prior to export to Matlab (version 2009a, Mathworks, Natwick, MA, USA) for post-processing.

3.4. Kinematics

During the 5-minute rest following collection of clinical measures, participants were fitted with the motion capture equipment (Fig. 1). A modified Helen-Hayes marker protocol was used (Kadaba et al., 1990) to place active infra-red markers bilaterally over the ASIS and PSIS on the pelvis, lateral femoral condyle, lateral malleolus, and on the outside of the shoe in locations to best represent the lateral calcaneus and fifth metatarsal head. Marker mounting wands were placed over the lateral femur, and at the level of the tibial tuberosity. Movement data was captured using four CodaMotion Cx1 sensor units (Charnwood Dynamics, Rotheley, Leicestershire, UK) sampling at 200 Hz.

4. Step up task

A step up task was used to assess the effects of the APFOS due to significant biomechanical differences between PFP populations and asymptomatic controls having been successfully identified during this task (Aminaka et al., 2011). Participants stepped up onto a Kistler force plate (Type 9281CA, Kistler Corporation, Switzerland) mounted on a wooden step (combined height 22 cm), in response to a verbal command. The dominant leg was always the lead leg, with all participants performing up to five practise trials to ensure the correct sequence. Participants were randomised into either APFOS—no APFOS or no APFOS—APFOS test groups by a coin toss. Data was collected during five separate repetitions for each test condition, with a 30 s sitting break between the two test conditions [Fig. 1]. Participants wore standardised neutral footwear (Asics Nimbus, Asics, Cheshire, UK) during all test conditions, with the APFOS (Vasyli easy fit, Vasyli, Essex, UK) [Fig. 3] placed directly into the heel of both left and right shoes when required. The prefabricated APFOS used has been designed to reduce rearfoot pronation through an inbuilt 6° varus posting in the heel.

5. Data analysis

The raw EMG was rectified and smoothed using a 0.02 s running median method. A time window 0.5 s prior to and 0.5 s post initial contact (determined from the first positive change in the force record exceeding a >10 N threshold) was setup, within which kinematic and EMG data was further analysed. Peak EMG amplitude values were extracted within this time window for each subject and averaged across the five trials for each test condition. Muscle onsets of GMed, VMO and VL were identified using a novel algorithm due to no previously identified methods being reported as optimal, and attempts to implement on our data being unsuccessful. A rise of EMG activity above a predetermined threshold that was maintained for a period of >30 ms was described as muscle onset. The threshold was calculated from the minimum of the means of all trials plus 10% of the range (maxima of means of all trials minus minima of means of all trials). The algorithm-determined muscle onset was imposed onto the EMG record to allow for visual confirmation. Negative values represent muscle onset prior to foot contact, and a positive value was subsequent to foot contact. Kinematic data across the three planes and two joints (hip and knee) were averaged across the five trials for each subject, and were extracted at four time points (−100 ms, 0 ms, +100 ms and +200 ms) either side of initial contact for further analysis between conditions.

6. Statistical analysis

Statistical analysis was performed using SPSS (version 18.0, SPSS inc., Chicago, IL). Between-condition comparisons were made using paired t-tests for GMed, VMO and VL onset times and peak amplitudes, for peak hip adduction, hip internal rotation angles, knee valgus and knee internal rotation angles. Bonferroni adjustment was not made for pair-wise comparisons to ensure potentially clinically meaningful findings were not missed due to stringent statistical correction (Perneger, 1998). Subsequent Spearman’s rank correlation analysis was calculated between the clinical measures and the change in biomechanical variables induced by the APFOS. The alpha level was set at 0.05.

7. Results

7.1. Hip and knee kinematics

A significant reduction in hip adduction angles of 1.56° was observed with APFOS application (P = 0.042) at 100 ms post initial contact.
(Fig. 4). A trend towards reduction in hip adduction angle was also observed at initial contact (1.19°, \(P = 0.065\)) and at 200 ms post initial contact (1.87°, \(P = 0.058\)), demonstrating a pattern of frontal plane movement modification (Fig. 3). A significant reduction in knee internal rotation angles of 1.3° was observed at initial contact (\(P = 0.043\)); however, no significant differences in hip transverse, or knee frontal plane movements were observed (Fig. 3, Table 1).

7.2. Muscle onset measures

No significant differences between test conditions were observed in the onset timing of GMed, VMO or VL (Table 1).

7.3. Muscle amplitude measures

No significant differences in normalised peak amplitudes of any muscles were seen between test conditions (Table 1).

7.4. Clinical measures

14 participants scored a normal (0 to 5) FPI score, two scored a pronated score (6 to 9) and one a highly supinated (−5 to −12) score. Mean KBAD range of motion was 50.2° ± 5.84°, with values ranging from 42° to 62° within the group.

7.5. Relationship of clinical measures to biomechanical changes induced by APFOS application

KBAD range of motion correlated significantly negatively with VL amplitude change (\(r = -0.53\), \(P = 0.019\)), and positively with hip adduction angle changes at initial contact and +200 ms (\(r = 0.52\) & 0.59, \(P = 0.027\) & 0.013 respectively). This demonstrated that less KBAD range of motion correlated with a greater reduction in hip adduction secondary to APFOS intervention. No significant correlations between muscle onsets, knee kinematics, and FPI scores were observed (Table 2).

8. Discussion

The purpose of this study was to investigate the effects of APFOS on biomechanical factors associated with PFP and its management. Findings indicated that during completion of a functional step up task, APFOS result in a significant reduction in hip adduction angles 100 ms post initial foot contact, and knee internal rotation angles at initial contact. A trend towards reduced hip adduction at initial contact and 200 ms post initial contact suggests a pattern of frontal plane motion control with the APFOS. Although changes observed were small in magnitude, it has been posited that these may be sufficient during repetitive activities to result in significant clinical change (Mills et al., 2010). Knee bent ankle dorsiflexion (KBAD) range of motion correlated significantly with the reduction in hip adduction angles observed, suggesting those with reduced KBAD range responded to the APFOS with greater reduction in peak hip adduction. The absence of a significant change in both muscle onsets and peak amplitudes at the hip and knee, gives some indication that the mechanism of APFOS is unlikely to be driven by changes in muscle activity.

Current literature indicates increased peak hip adduction and internal rotation angles at the hip during functional tasks are commonly demonstrated in individuals with PFP (Aminaka et al., 2011; Willson et al., 2011). Findings from this study suggest APFOS have a positive...
impact on these lower limb kinematic characteristics, providing a possible mechanistic explanation for their reported efficacy with PFP populations (Barton et al., 2010; Collins et al., 2008). Specifically, reduced peak hip adduction with the APFOS during repeated functional activities may reduce PF stress and subsequent persistence of pain (Tiberio, 1987). Tiberio (1987) proposed that excessive sub-talar joint pronation resulted in delayed external tibial rotation during knee extension leading to a compensatory increase in internal femoral rotation and resultant increased retropatellar contact pressure and duration. Potential changes evoked at the sub-talar joint through the addition of an APFOS have the capacity to result in the kinematic changes throughout the kinetic chain that were observed within this study.

The clinical relevance of isolated reduction in knee internal rotation resulting from APFOS intervention is however questionable, due largely to the inconsistent findings linking sub-talar joint pronation to increased internal tibial rotation within PFP populations (Reischl et al., 1999) and the magnitude of the change observed (1.3°). In addition, in vitro studies have reported that increased internal tibial rotation has no influence on patellofemoral joint contact area or pressures (Lee et al., 2003). The observed kinematic changes at the knee however, may support that the mechanism of APFOS effect is not exclusively at the knee, but throughout the kinetic chain. Further research evaluating the effects of APFOS on PFJ forces combined with hip muscle function and kinematics in a PFP population is needed to confirm this.

The correlation observed between KBAD range of motion, hip kinematics, and EMG variables, demonstrates a significant trend for individuals who have reduced KBAD to respond with a greater reduction in peak hip adduction and reduced VL peak amplitude following an APFOS intervention. Reduced ankle range of motion has been reported previously within PFP populations to be a significant predictor of APFOS intervention success (Barton et al., 2011), these findings could provide a potential mechanistic explanation for this clinically meaningful outcome.

The observed lack of significant change in GMd, VMO and VL onset times between conditions were consistent with previous reports during a single leg perturbation task (Rose et al., 2002). It has been proposed that this lack of change may be due to the short term nature of the intervention, without sufficient time for alterations in activation patterns to occur (Rose et al., 2002). Habituation has been demonstrated to induce earlier onset times within the erector spinae muscle groups (Barton et al., 2009), and increase peroneus longus EMG amplitude (Murley and Bird, 2006), indicating the potential for significant EMG changes following a prolonged period using orthoses intervention. Furthermore, it is possible that within asymptomatic participants, less likely to exhibit altered quadriceps muscle onsets compared to PFP populations (Chester et al., 2008), that the orthoses cannot generate a change to these activation patterns. Further research within PFP populations is clearly warranted to explore muscle activation mechanisms further.

The difference observed between the findings of this study and another that reported significant GMd, VMO and VL amplitude changes with the addition of an orthoses, was that participants had previously been completing a maximal single leg squatting task, likely to result in a greater muscle demand and therefore EMG signal (Hertel et al., 2005). With amplitude changes associated with foot orthoses considered to be highly variable between individuals (Nigg et al., 1999), in combination with completion of differing tasks in varying degrees of orthotic rear foot posting, a lack of a consistent finding between studies is not unexpected.

The primary limitation of this study is that not using Bonferroni corrections in an attempt to reduce the possible risk of Type 2 error resulted in an increased risk of Type 1 error. Additionally, with a lack of comparative PFP participants it must be with caution that these results are extrapolated to PFP populations. However, it is possible that within PFP populations who have been reported to demonstrate greater peak hip adduction and internal rotation angles (Aminaka et al., 2011; Willson et al., 2011), that even greater changes might be observed. Further, it may be that sub-groups of subjects exist who may exhibit varying responses to APFOS.

Unmodified prefabricated APFOS were used in this study for practical reasons. However, this is unlikely to fully replicate clinical practice where APFOS are frequently moulded and/or customised with varying wedging. Further studies tailoring the APFOS to the individual might demonstrate greater biomechanical changes. Additionally, the provision of a foot orthoses that limits pronation may restrict mid-foot compensation for a lack of ankle dorsi-flexion range—the long-term consequences of which need to be borne in mind during clinical practise.

### Table 1

<table>
<thead>
<tr>
<th>Time</th>
<th>Hip angle (°)</th>
<th>Knee angle (°)</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Coronal plane</td>
<td>Transverse plane</td>
</tr>
<tr>
<td></td>
<td>MD</td>
<td>CI</td>
</tr>
<tr>
<td>−100 ms</td>
<td>0.72</td>
<td>−0.5 to 2.0</td>
</tr>
<tr>
<td>IC</td>
<td>1.19</td>
<td>−0.08 to 2.5</td>
</tr>
<tr>
<td>+100 ms</td>
<td>1.56</td>
<td>0.06 to 3.05</td>
</tr>
<tr>
<td>+200 ms</td>
<td>1.87</td>
<td>−0.07 to 3.82</td>
</tr>
</tbody>
</table>

Key: −100 ms = 100 ms before initial contact with the force plate; IC = initial contact with force plate; +100 ms = 100 ms after initial contact with force plate; +200 ms = 200 ms after initial contact with the force plate; PFI = foot posture index; KBAD RoM = knee bent ankle dorsiflexion range of motion.

### Table 2

<table>
<thead>
<tr>
<th>Biomechanical variables</th>
<th>Clinical measures</th>
<th>FPI</th>
<th>r value</th>
<th>P value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Onset</td>
<td></td>
<td></td>
<td></td>
<td></td>
</tr>
<tr>
<td>GMd</td>
<td>−0.29</td>
<td>0.251</td>
<td>0.2</td>
<td>0.429</td>
</tr>
<tr>
<td>VMO</td>
<td>−0.16</td>
<td>0.531</td>
<td>0.14</td>
<td>0.583</td>
</tr>
<tr>
<td>VL</td>
<td>−0.29</td>
<td>0.242</td>
<td>0.05</td>
<td>0.841</td>
</tr>
<tr>
<td>−100 ms</td>
<td>0.23</td>
<td>0.52</td>
<td>0.06</td>
<td>0.800</td>
</tr>
<tr>
<td>0.23</td>
<td>0.52</td>
<td>0.06</td>
<td>0.800</td>
<td></td>
</tr>
<tr>
<td>0.23</td>
<td>0.52</td>
<td>0.06</td>
<td>0.800</td>
<td></td>
</tr>
<tr>
<td>−100 ms</td>
<td>0.24</td>
<td>0.243</td>
<td>0.55</td>
<td>0.190</td>
</tr>
<tr>
<td>0.243</td>
<td>0.55</td>
<td>0.190</td>
<td></td>
<td></td>
</tr>
<tr>
<td>0.243</td>
<td>0.55</td>
<td>0.190</td>
<td></td>
<td></td>
</tr>
<tr>
<td>Hip (x)</td>
<td>−0.37</td>
<td>0.147</td>
<td>0.31</td>
<td>0.228</td>
</tr>
<tr>
<td>IC</td>
<td>−0.12</td>
<td>0.624</td>
<td>0.02</td>
<td>0.948</td>
</tr>
<tr>
<td>−100 ms</td>
<td>0.21</td>
<td>0.395</td>
<td>0.1</td>
<td>0.698</td>
</tr>
<tr>
<td>0.215</td>
<td>0.395</td>
<td>0.1</td>
<td>0.698</td>
<td></td>
</tr>
<tr>
<td>0.215</td>
<td>0.395</td>
<td>0.1</td>
<td>0.698</td>
<td></td>
</tr>
<tr>
<td>−100 ms</td>
<td>0.16</td>
<td>0.544</td>
<td>0.59</td>
<td>0.013*</td>
</tr>
<tr>
<td>0.16</td>
<td>0.544</td>
<td>0.59</td>
<td>0.013*</td>
<td></td>
</tr>
<tr>
<td>0.16</td>
<td>0.544</td>
<td>0.59</td>
<td>0.013*</td>
<td></td>
</tr>
<tr>
<td>Knee (x)</td>
<td>0.33</td>
<td>0.181</td>
<td>0.15</td>
<td>0.261</td>
</tr>
<tr>
<td>IC</td>
<td>−0.38</td>
<td>0.123</td>
<td>0.38</td>
<td>0.116</td>
</tr>
<tr>
<td>−100 ms</td>
<td>−0.19</td>
<td>0.459</td>
<td>0.08</td>
<td>0.756</td>
</tr>
<tr>
<td>0.225</td>
<td>0.459</td>
<td>0.08</td>
<td>0.756</td>
<td></td>
</tr>
<tr>
<td>0.225</td>
<td>0.459</td>
<td>0.08</td>
<td>0.756</td>
<td></td>
</tr>
<tr>
<td>−100 ms</td>
<td>−0.12</td>
<td>0.172</td>
<td>0.23</td>
<td>0.368</td>
</tr>
<tr>
<td>0.027</td>
<td>0.172</td>
<td>0.23</td>
<td>0.368</td>
<td></td>
</tr>
<tr>
<td>0.027</td>
<td>0.172</td>
<td>0.23</td>
<td>0.368</td>
<td></td>
</tr>
<tr>
<td>Knee (z)</td>
<td>0.02</td>
<td>0.928</td>
<td>0.1</td>
<td>0.692</td>
</tr>
<tr>
<td>IC</td>
<td>0.02</td>
<td>0.928</td>
<td>0.1</td>
<td>0.692</td>
</tr>
<tr>
<td>−100 ms</td>
<td>0.13</td>
<td>0.607</td>
<td>0.13</td>
<td>0.603</td>
</tr>
<tr>
<td>0.03</td>
<td>0.607</td>
<td>0.13</td>
<td>0.603</td>
<td></td>
</tr>
<tr>
<td>0.03</td>
<td>0.607</td>
<td>0.13</td>
<td>0.603</td>
<td></td>
</tr>
</tbody>
</table>

Key: −100 ms = 100 ms before initial contact with the force plate; IC = initial contact with force plate; +100 ms = 100 ms after initial contact with force plate; +200 ms = 200 ms after initial contact with the force plate; PFI = foot posture index; KBAD RoM = knee bent ankle dorsiflexion range of motion.
Step up time was not standardised between individuals in this study in order to ensure that individualised movement pattern was captured, however, the consequence of this may have been that the variability of the speed completing the task may have altered aspects of the EMG and kinematic variables. Using initial contact to stack individual trials, and with comparisons being made within subjects likely to adopt a similar movement pattern between conditions is expected to be sufficient to minimise this problem.

It is important to acknowledge, that given the dearth of current literature linking EMG and kinematic variables to the effects of APFOS interventions, this study offers potential mechanisms underlying observed clinical efficacy. It demonstrates the potential for prefabricated APFOS to result in significantly reduced hip adduction and knee internal rotation angles, characteristics evident within PFP populations. An exploration into the effects of APFOS in PFP patients requires further work.

9. Conclusion

Results from this study indicate that prefabricated APFOS can reduce hip adduction and knee internal rotation during a functional step up task. Increased KBAD correlated significantly with the changes in hip adduction angles. These biomechanical observations are highly relevant to patient groups, such as the high number of PFP sufferers, and may partially underlie the mechanism behind established therapeutic effects observed from APFOS interventions. Further research in patients with pathology is warranted to clarify mechanisms of successful treatment mechanisms and therefore individual predictors of treatment efficacy.

Acknowledgements

The authors acknowledge Vasyli UK for the supply of the APFOS device used in this study and Ascis UK for the supply of standardised footwear.

References


