Mechanical effects of medial and lateral wedged orthoses during running.

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Abstract

OBJECTIVE: The aim of the current investigation was to examine the effects of orthoses with 5° medial and lateral wedges on knee joint kinetics during the stance phase of running.

DESIGN: Repeated measures

SETTING: Laboratory

PARTICIPANTS: Twelve recreational runners

OUTCOME MEASUREMENTS: Twelve male participants ran over a force platform at 4.0 m/s in three different conditions (medial orthotic, lateral orthotic and no-orthotic). Lower limb kinematics were collected using an 8-camera motion capture system allowing knee kinetics to be quantified using a musculoskeletal modelling approach. Differences in knee joint kinetics between orthotic conditions were examined using one-way repeated measures ANOVA.

RESULTS: The results showed that peak patellofemoral force was significantly increased in the medial (31.81 N/kg) and lateral (31.29 N/kg) wedged orthoses, in comparison to the no-orthotic (29.61 N/kg) condition. In addition, the peak knee adduction moment was significantly increased in the medial (1.10 Nm/kg) orthoses, in comparison to the lateral (0.87 Nm/kg) condition.

CONCLUSIONS: The results from this study indicate that lateral orthoses may be effective in attenuating runners risk from medial tibiofemoral compartment OA, but that wedged orthoses may enhance their risk from patellofemoral pain.

Introduction

Although distance running is associated with a plethora of physiological benefits (Lee et al., 2014), it is also linked with a very high rate of overuse injuries (Taunton et al., 2002), with an occurrence rate of up to 70 % per year of training (Van Gent et al., 2007). The knee joint is the musculoskeletal site that is most likely to experience an overuse injury (van Gent et al., 2007). Specifically, patellofemoral pain and pain secondary to knee osteoarthritis are common complaints reported by runners (Taunton et al., 2002).

Patellofemoral pain syndrome is regarded as the most common overuse injury in runners (Taunton et al., 2002). Pain symptoms present clinically as isolated pain at the anterior aspect of the patella (Ho et al., 2012). As knee flexion proceeds from full extension, the pull of the quadriceps and patellar tendon becomes increasingly oblique, compressing the patella against the femur and generating the patellofemoral joint reaction force (Trepczynski et al., 2012). The dynamics of the knee in the sagittal plane may have a prominent effect on the patellofemoral joint, and a model to estimate the patellofemoral joint reaction force has previously been developed (Ward & Powers, 2004). Elevated patellofemoral joint stress, which is a reflection of the patellofemoral joint reaction force divided by the patellofemoral contact area, is the most commonly accepted aetiological factor in the development of patellofemoral pain syndrome (Farrokhi et al., 2011). Excessive rearfoot eversion/ tibial internal rotation during the stance phase, necessitates greater hip internal rotation and adduction (Barton et al., 2011). These mechanisms are thought enhance patellofemoral stress, owing to a reduced joint contact area (Tiberio, 1987). Patellofemoral pain symptoms can cause training restrictions (Waryasz & McDermott, 2008), and pain symptoms associated with patellofemoral disorders can persist for many years (Collins et al., 2013). Importantly, 45-64% of individuals with patellofemoral osteoarthritis (OA) report patellofemoral pain symptoms during adolescence or early adulthood (Crossley, 2014).

Degenerative tibiofemoral pathologies are also common in runners; accounting for as many as many as 16.8 % of all chronic knee injuries (Taunton et al., 2002). The causes of tibiofemoral chronic pathologies relate to the magnitude of compressive loading at the joint (Morgenroth et al., 2014), which is considered to be the mechanical parameter most strongly associated with the onset and progression of knee OA. The medial aspect of the tibiofemoral joint is known to be significantly more prone to osteoarthritic degeneration than the lateral compartment (Wise et al., 2012). In vivo analyses have shown that compressive loading experienced by the medial aspect of the tibiofemoral joint is correlated positively with the magnitude of the knee adduction moment (KAM) (Zhao et al., 2007; Kutzner et al., 2013). Therefore, the KAM is frequently utilized as a pseudo measure of medial tibiofmeoral contact loading (Birmingham et al., 2007), and the peak KAM as well as the slope of the KAM have been cited as important predictors of radiographic knee OA (Miyazaki et al., 2002; Morgenroth et al., 2014).

Given their prevalence and debilitating nature, numerous strategies have been investigated in clinical research in an attempt to attenuate the risk of knee pathologies in runners. Foot orthoses are one of the most popular conservative options for the prevention/ treatment of knee pathologies in runners (Heiderscheit et al., 2001). For patellofemoral pain symptoms, foot orthoses have importantly, been shown to be successful in improving pain symptoms and function (Collins et al., 2008; Barton et al., 2011).

In addition to traditional foot orthoses, wedged orthoses that are built up along either the medial or lateral edges have become common in recent years (Aminian et al., 2014). Medially orientated foot orthoses are often utilized to reduce lower extremity biomechanics linked to increases in patellofemoral stress by attenuating rearfoot eversion/ tibial internal rotation during the stance phase (Boldt et al., 2013). However, using a sagittal plane model to estimate the patellofemoral loading, Almonroeder et al., (2015), showed that prefabricated foot orthoses with 5° of medial rearfoot wedging significantly increased peak patellofemoral stress compared to running without orthoses. Similarly, laterally wedged orthoses have also been advocated as a mechanism that may reduce the magnitude of the KAM and thus the loads experienced by the medial compartment of the tibiofemoral joint (Yamaguchi et al., 2015). Lewinson et al., (2013) who investigated the effects of 3, 6, and 9 mm medial/ lateral wedged footwear on coronal plane knee moments during running, showed that laterally wedged running footwear were associated with significant reductions in the peak KAM. Nigg et al., (2003) examined the effects of medial, lateral and neutral shoe inserts on knee joint moments during heel-toe running. Compared with the neutral insert condition, the maximal external knee rotation moment was found to be significantly greater in the full medial insert condition.

However, whilst the effects of foot orthoses on the biomechanics the knee joint during gait have been examined previously, there has yet to be any investigation which has collectively explored the effects of medial and lateral orthoses on the kinetics of the patellofemoral and tibiofemoral joints during running. Therefore, the aim of the current investigation was to examine the effects of orthoses with a 5° medial and lateral wedge on knee joint kinetics during the stance phase of running. A clinical investigation of this nature may provide insight into the potential efficacy of wedged foot orthoses for the prevention of knee pathologies in runners. The current investigation tests the hypotheses that medial orthoses will reduce patellofemoral joint loading and lateral orthoses will reduce the magnitude of the KAM during the stance phase of running.

Methods

Participants

Twelve male runners (age 26.23 ± 5.76 years, height 1.79 ± 0.11 cm and body mass 73.22 ± 6.87 kg) volunteered to take part in this study. The sample was based on previous analyses, which have examined the effects of wedged orthoses on lower extremity kinetics during running (Almonroeder et al., 2015; Lewinson et al., 2013). All participants identified as recreational runners, who trained a minimum of 3 times/week completing a minimum of 35 km/week. All participants were also free from knee pathology at the time of data collection and had not previously had any knee surgery. The participants provided written informed consent in accordance with the principles outlined in the Declaration of Helsinki. The procedure utilized for this investigation was approved by the University of Central Lancashire, Science, Technology, Engineering and Mathematics, ethical committee (REF 357).

Orthoses

Commercially available full-length orthoses (Slimflex Simple, High Density, Full Length, Algeos UK) were examined in the current investigation (Figure 1-2). The orthoses were made from ethylene-vinyl acetate with a shore A rating of 65 and had a heel thickness of 11 mm including the additional wedge. The orthoses were able to be modified to either a 5˚ varus or valgus configuration which in two separate components spanned the full length of the device (Figure 1-2). To ensure consistency each participant wore the same footwear (Asics, Patriot 6). The experimental footwear had a mean mass of 0.265kg, heel thickness of 22mm and heel drop of 10mm. The order that participants ran in each orthotic condition was counterbalanced.

@@@ Figure 1 near here @@@

@@@ Figure 2 near here @@@

Procedure

Participants ran at 4.0 m/s (±5%), striking an embedded piezoelectric force platform (Kistler, Kistler Instruments Ltd., Alton, Hampshire) with their right (dominant) foot (Sinclair et al., 2014a). Running velocity was monitored using infrared timing gates (Newtest, Oy Koulukatu, Finland). The stance phase was delineated as the duration over which 20 N or greater of vertical force was applied to the force platform (Sinclair et al., 2011). Runners completed five successful trials in each orthotic condition (medial, lateral and no-orthotic).

Kinematic data was captured at 250 Hz via an eight camera motion analysis system (Qualisys Medical AB, Goteburg, Sweden). Kinematics and ground reaction forces data were synchronously collected. Dynamic calibration of the motion capture system was performed before each data collection session.

Lower extremity segments were modelled in 6 degrees of freedom using the calibrated anatomical systems technique (Cappozzo et al., 1995). To define the segment co-ordinate axes of the shank and thigh, retroreflective markers were placed bilaterally onto the medial and lateral malleoli, medial and lateral epicondyles of the femur. To define the pelvis segment further markers were posited onto the anterior (ASIS) and posterior (PSIS) superior iliac spines. Carbon fiber tracking clusters were positioned onto the shank and thigh segments. The pelvis was tracked using the ASIS and PSIS markers. The centre of the knee joint was delineated as the mid-point between the femoral epicondyle markers (Sinclair et al., 2015a), whereas the hip joint centre was obtained using the positions of the ASIS markers (Sinclair et al., 2014b). Static calibration trials were obtained allowing for the anatomical markers to be referenced in relation to the tracking markers/ clusters. The Z (transverse) axis was oriented vertically from the distal segment end to the proximal segment end. The Y (coronal) axis was oriented in the segment from posterior to anterior. Finally, the X (sagittal) axis orientation was determined using the right hand rule and was oriented from medial to lateral.

Processing

Dynamic trials were digitized using Qualisys Track Manager in order to identify anatomical and tracking markers then exported as C3D files to Visual 3D (C-Motion, Germantown, MD, USA). Ground reaction force and kinematic data were smoothed using cut-off frequencies of 50 and 12 Hz respectively with a low-pass Butterworth 4th order zero-lag filter. Net joint moments were calculated using Newton-Euler inverse dynamics.

A previously utilized mathematical model was used to estimate patellofemoral contact force and patellofemoral contact stress during the stance phase of running (Ward & Powers, 2004). This model has been utilized previously to successfully resolve differences in contact force and stress when wearing different footwear (Bonacci et al., 2013; Sinclair, 2014, Sinclair et al., 2016) and between orthoses (Sinclair et al., 2015b) during running. Patellofemoral contact force was estimated as a function of the knee flexion angle and knee flexion moment according to the biomechanical model described by Ho et al., (2012). Firstly, an effective moment arm of the quadriceps muscle was calculated as a function of the knee flexion angle using a non-linear equation, which is based on cadaveric data presented by van Eijden et al., (1986):

$$\textbf{Quadriceps moment arm} = \mathbf{0.00008 * knee\ flexion\ angle^{3} - 0.013 * knee\ flexion\ angle^{2} + 0.28 * knee\ flexion\ angle + 0.046}$$

Quadriceps force was then estimated using the below formula:

Quadriceps force = knee flexion moment / quadriceps moment arm

Patellofemoral contact force was estimated using the quadriceps force and a constant:

Patellofemoral contact force = quadriceps force * constant

The constant was described in relation to the knee flexion angle using a curve fitting technique based on the non-linear equation described by Eijden et al., (1986):

$$\mathbf{constant = (0.462 + 0.00147 * knee\ flexion\ angle^{2} – 0.0000384 * knee\ flexion\ angle^{2}) / (1 – 0.0162 * knee\ flexion\ angle + 0.000155 * knee\ flexion\ angle^{2} – 0.000000698 * knee\ flexion\ angle^{3})}$$

Contact stress (MPa) was estimated as a function of the contact force divided by the patellofemoral contact area. The contact area was described in accordance with the Ho et al., (2012) recommendations by fitting a 2^{nd} order polynomial curve to the data of Powers et al., (1998), which documented patellofemoral contact areas at varying levels of knee flexion.

Patellofemoral contact stress = patellofemoral contact force / contact area

Knee loading was examined through extraction of the peak KAM, peak patellofemoral contact force and peak patellofemoral contact stress. Patellofemoral contact force parameters were normalized (N/kg) by dividing the net values by body mass. KAM load rate (Nm/kg/s) was also calculated by dividing the peak KAM by the time taken. Finally, we calculated the total patellofemoral contact force impulse (N/kg·s) using a trapezoidal function.

Statistical Analyses

Means and standard deviations were calculated for each outcome measure for all orthotic conditions. Differences in knee kinetic parameters between orthotic conditions were examined using one-way repeated measures ANOVAs, with significance accepted at the $P \leq 0.05$ level. Post-hoc pairwise comparisons with a Bonferroni adjustment to control type I error, were conducted on all significant main effects. Effect sizes were conducted for each main effect and for all significant pairwise comparisons, using partial eta^2 ($p\eta^2$). Effect sizes were contextualized using the following guidelines; small = 0.01, medium = 0.06 and large = 0.14 (Cohen, 1988). The data was screened for normality using a Shapiro-Wilk, which confirmed that the normality assumption was met. All statistical analyses were conducted using SPSS v23.0 (SPSS Inc., Chicago, USA).

Results

Figure 3 and table 1 present the differences in knee kinetic parameters as a function of different orthotic configurations.

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Patellofemoral kinetics

A significant main effect was noted for peak patellofemoral contact force ($P<0.05$, $p\eta^2 = 0.29$). Post-hoc pairwise comparisons showed that peak patellofemoral contact force was significantly greater in the lateral ($P=0.041$, $p\eta^2 = 0.31$) and medial ($P=0.045$, $p\eta^2 = 0.31$) configurations, in relation to the no-orthoses condition (Figure 3a; Table 1). However, there was no main effect for peak patellofemoral stress ($P<0.05$, $p\eta^2 = 0.17$, Figure 3b; Table 1).

Finally, a significant main effect ($P<0.05$, $p\eta^2=0.37$) was noted for patellofemoral impulse. Post-hoc pairwise comparisons showed that patellofemoral impulse was significantly greater in the lateral ($P=0.012$, $p\eta^2=0.45$) and medial ($P=0.027$, $p\eta^2=0.37$) configurations, in relation to the no-orthoses condition (Table 1).

Knee adduction moment parameters

A significant main effect ($P<0.05$, $p\eta^2 = 0.31$) was observed for the magnitude of peak KAM. Post-hoc pairwise comparisons showed that peak KAM was significantly larger in the medial orthoses in relation to the lateral orthoses ($P=0.03$, $p\eta^2=0.35$) (Figure 3c; Table 1). There was however, no main effect for the KAM load rate ($P<0.05$, $p\eta^2 = 0.12$, Table 1).

Discussion

The aim of the present study was to examine the influence of orthoses with 5° medial and lateral wedges on knee joint kinetics during the stance phase of running. To the authors knowledge this represents the first investigation to collectively explore the effects of medial and lateral orthoses on the kinetics of the patellofemoral and tibiofemoral joints during running. The findings from this investigation provide partial support for the hypotheses in that lateral orthoses significantly reduced the magnitude of the peak KAM, but both medial and lateral orthoses significantly increased patellofemoral joint loading during the stance phase of running.

Patellofemoral pain is widely acknowledged as the most common overuse running pathology (Taunton et al., 2002). The current investigation showed that patellofemoral loading parameters were significantly greater when running in the medial and lateral orthotic modalities compared to running without any orthotic intervention. This observation supports the findings of Almonroeder et al., (2015) who observed increases in patellofemoral loading when running in medial orthoses, although increases patellofemoral joint kinetics when with lateral orthoses the has not been shown previously. It is important however that the statistical observations at the patellofemoral joint be contextualized in relation to the mean difference between conditions. The mean differences between conditions were relatively small, thus it is unknown whether the statistical observations are also clinically significant. Nonetheless, this finding may still be important regarding the initiation and progression of patellofemoral pain, as the patellofemoral pain symptoms are mediated through excessive patellofemoral joint loading (Farrokhi et al., 2011). Therefore, current study indicates that running with wedged foot orthoses as a prophylactic modality for patellofemoral pain may not be justified, although further longitudinal analyses are required before this can be clinically substantiated.

It is proposed that the mechanism responsible for the increases in patellofemoral loading in the wedged orthotic conditions was an enhanced knee flexion moment. Similar to the proposition offered by Almonroeder et al., (2015) the additional heel elevation (11 mm) provided by the orthotic conditions may have influenced the vector orientation of the ground reaction force so that the magnitude of the knee flexion moment, a key input parameter into the patellofemoral model was increased. Previous trials have shown that foot orthoses served to improve patellofemoral pain symptoms (Collins et al., 2008; Barton et al., 2011); the findings from the current study indicate that the clinical improvements in pain symptoms may not have been mediated through alterations in sagittal plane knee mechanics.

In addition, the current investigation also showed the peak KAM was significantly reduced in the lateral orthotic condition in relation to the medial and no-orthotic conditions. This agrees with the observations of Lewinson et al., (2013) who showed that laterally wedged running footwear significantly reduced the peak KAM. It is proposed that this observation is mediated by the effects of the lateral orthoses themselves by attenuating the magnitude of the ground reaction force moment arm about the knee joint centre. The peak KAM is considered an effective pseudo measure of compressive medial compartment loading (Birmingham et al., 2007), and is believed to be an important biomechanical predictor of the initiation and progression of radiographic knee OA (Miyazaki et al., 2002). Again, it is important to contextualize the mean differences in peak KAM between the medial and lateral orthoses which was relatively small. As such it is not known whether the statistical changes in the

KAM are clinically significant. It appears that lateral orthoses may be able to attenuate the risk from the kinetic parameters linked to the aetiology of medial compartment knee OA in runners. This therefore presents an interesting paradox in that lateral orthoses may attenuate biomechanical risk factors in those susceptible to medial knee OA, yet appear to increase the mechanisms linked to the aetiology of patellofemoral pain. This is a clear avenue for future clinical research, to determine the long-term effects of lateral orthoses in runners.

A potential limitation of the current investigation is that it examines healthy male runners who habitually did not wear orthotics. Firstly, as female runners are known to be more susceptible to overuse knee injuries (Ivković et al., 2007), and secondly as it is not possible to determine if the findings are generalizable to runners with existing patellofemoral pain or medial compartment knee OA. Future, analyses will help to determine the clinical efficacy of wedged orthoses as treatment modalities for runners of both sexes, with existing chronic knee injuries. A further potential drawback is the method by which patellofemoral stress was quantified. Sagittal knee mechanics as input parameters into the mathematical model do not account for the effects of coronal/ transverse plane knee kinematics on the patellofemoral joint contact area. Further advancements in musculoskeletal research are required to provide a three-dimensional model of the patellofemoral joint contact area allowing joint stress to be calculated more accurately.

In conclusion, despite the fact that the biomechanical effects of foot orthoses have been examined previously, current knowledge with regards to the effects of medial and lateral orthoses on the loads experienced by the patellofemoral and tibiofemoral joints during running is limited. This study therefore adds to the current literature in the field of clinical biomechanics by giving a comprehensive comparative examination of patellofemoral and tibiofemoral loading parameters during the stance phase of running in medial and lateral orthoses. The current investigation importantly showed that lateral orthoses attenuated the magnitude of the KAM but that wearing wedged orthoses increased patellofemoral loading parameters. The results from this study indicate that lateral orthoses may be effective in attenuating runners risk from medial tibiofemoral compartment OA, but that wedges orthoses may enhance their risk from patellofemoral pain.

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Table 1: Knee kinetics (Means, standard deviations and 95% confidence intervals) as a function of the different orthotic conditions.

	No-orthotic		**Medial**		**Lateral**		
	Mean	***SD***	***Mean***	***SD***	***Mean***	***SD***	
Peak patellofemoral contact force (N/kg)	29.61***AB***	9.35	31.81	9.65	31.29	9.04	*
Patellofemoral impulse (N/kg·s)	2.44 ***AB***	1.1	2.82	1.37	2.7	1.25	*
Peak patellofemoral stress (MPa)	8.81	2.68	9.33	2.71	9.37	2.54	
Peak KAM (Nm/kg)	0.93	0.41	1.1	0.4	0.87 ***A***	0.34	*
KAM load rate (Nm/kg/s)	25.2	17.89	24.03	16.55	24.72	16.57	

Key: * = significant main effect

A = significantly different from Medial orthosis

B = significantly different from Lateral orthosis

List of figures

Figure 1: Figure 1: Rear image of the experimental orthoses (a. = 5° medial configuration and b. = 5° lateral configuration).

Figure 2: Medial and lateral images of the experimental orthoses (a. = lateral view of 5° lateral configuration, b. = lateral view of 5° medial configuration, c. = medial view of 5° lateral configuration, d. = medial view of 5° medial configuration).

Figure 3: Knee joint kinetics as a function of different orthotic conditions (a. = patellofemoral force, b. = patellofemoral stress, c. = knee adduction moment) (Black = medial, dot = lateral, grey = no-orthotic).