EFFECTS OF MEDIAL AND LATERAL WEDGED ORTHOSES ON KNEE AND ANKLE JOINT LOADING IN FEMALE RUNNERS.

Keywords: Biomechanics; orthoses; kinetics; running.

Abstract

The aim of the current investigation was to examine the effects of orthoses with a 5° medial and lateral wedge on knee and ankle joint kinetics in female runners. Twelve healthy female runners ran at 3.5 m/s over a force platform in three conditions (medial, lateral and no-orthotic). Lower extremity kinematics were measured using an 8-camera motion capture system, which allowed knee and ankle loading to be explored using a musculoskeletal modelling approach. The peak Achilles tendon force was significantly larger in the no-orthotic condition (5.34 BW) compared to the lateral orthosis (5.03 BW). The peak patellofemoral stress was significantly larger in the medial orthosis (7.32 MPa) compared to the no-orthotic (7.02 MPa) condition. Finally, the peak knee adduction moment was significantly larger in the medial condition (1.14 Nm/kg) compared to the lateral (0.99 Nm/kg) orthosis. The findings from the current investigation indicate that lateral orthoses may be effective in attenuating risk from medial tibiofemoral osteoarthritis and Achilles tendinopathy, but medial wedge orthoses may increase the risk from patellofemoral pain in female runners.

Introduction

Running is linked to a high incidence of overuse injuries (Taunton et al., 2002; Hreljac, 2004), with an occurrence rate of up to 70% per year (Van Gent et al., 2007). The knee and ankle joints have been demonstrated as the most commonly injured musculoskeletal sites (van Gent et al., 2007). Importantly female runners are renowned for being at increased risk from chronic injuries in relation to males (Taunton et al., 2002).

Patellofemoral pain is the most common chronic injury encountered in sports medicine (Crossley, 2014), characterized by pain at or anterior to the patella exacerbated by cyclic physical activities such as running that frequently load the patellofemoral joint (Crossley et al., 2016). Pain symptoms typically persist for many years (Collins et al., 2013), and force many runners to mediate or even cease their training (Waryasz & McDermott, 2008). The peak patellofemoral joint stress; a manifestation of the patellofemoral joint reaction force divided by the patellofemoral contact area, is widely regarded as the most prominent biomechanical mechanism linked to the aetiology of patellofemoral pain syndrome (Farrokhi et al., 2011). Importantly, a recent systematic review has shown that there may be a link between patellofemoral pain in younger adults and subsequent osteoarthritis (OA) at this joint (Thomas et al., 2013).

Furthermore, chronic tibiofemoral pathologies are also common running injuries, and associated with up to 16.8% of all knee injuries (Taunton et al., 2002). The medial aspect of the knee is significantly more susceptible to injury 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 has been cited as an important predictor of radiographic knee OA (Miyazaki et al., 2002; Morgenroth et al., 2014).

Finally, Achilles tendinopathies are also frequently occurring chronic musculoskeletal disorders in runners, accounting for approximately 8–15% of all injuries (Van Ginckel et al., 2009). Although the Achilles is regarded as the strongest tendon in the body, it is the most common site of tendinous injury (Rice & Patel, 2017). During running the Achilles tendon experiences forces up to 7 BW (Almondroeder et al., 2013). Excessive cyclic forces experienced by the tendon during activities such as running are is regarded as the main pathological stimulus for the initiation of Achilles tendinopathy (Abate et al., 2009). With repeated high and insufficient time for repair, the reparative capability of the tendon is exceeded breaking the cross-links and causing degeneration of the tendons collagen fibrils (Cook & Purdam, 2009).

Taking into account the high incidence of running injuries, and the debilitating nature of chronic pathologies, a range of preventative mechanisms have been explored in biomechanical literature in order to attenuate the risk from injury in runners. Foot orthoses are one of the most commonly utilized modalities for the prevention/ treatment of running injuries (Bonanno et al., 2017). Foot orthoses are available in both medial and lateral configurations, which are utilized in order to specifically modify the alignment of the lower extremities and redistribute the loads experienced at the lower body joints (Liu & Zhang, 2013). The effects of medial/ lateral orthoses on the biomechanics the lower extremities have been examined previously, however they have habitually been examined during walking in pathological patients (Pham et al., 2004; Rubin et al, 2005; Baker et al, 2007; Barrios & Davis, 2010; Bennell et al., 2010; Rafianee & Karimi, 2012; Barrios et al., 2013) and there is only limited information concerning their effects during running.

Boldt et al., (2013) examined the effects of 6° medially wedged orthoses on the biomechanics of the hip and knee joint in female runners with and without patellofemoral pain. Their findings showed in both groups, that the peak KAM increased and the hip adduction excursion decreased when wearing the medial orthoses. Almonroeder et al., (2015), examined the effects of prefabricated foot orthoses with 5° of medial wedging in female runners. Medial orthoses significantly increased peak patellofemoral stress in comparison to running without orthoses. Lewinson et al., (2013) who explored the influence of 3, 6, and 9 mm medial and lateral wedged footwear on the KAM in males, showed that laterally wedged running footwear were associated with significant reductions in the peak KAM. Sinclair, (2018) studied the effects of 5° medial and lateral orthoses on knee joint loading in male runners. Their findings showed that patellofemoral loading was significantly increased in the medial and lateral orthoses compared to no-orthoses and the peak KAM was significantly increased in the medial compared to the lateral orthoses. Nigg et al., (2003) examined the effects of medial, lateral and neutral shoe inserts on knee joint moments during heel-toe running in males. Compared with the neutral insert condition, the maximal external knee rotation moment was found to be significantly greater in the medial insert condition. Starbuck et al., (2017) examined the effects of an off-the-shelf lateral wedge orthotic on knee loading in a mixed sample of runners. Their results showed that the orthoses did not statistically influence knee loading parameters during the stance phase. Using an in-vitro analysis Kogler et al., (1999) investigated the influence of medial and lateral orthotic wedges on loading of the plantar aponeurosis. Their findings showed that wedging under the lateral aspect of the forefoot decreased strain in the plantar aponeurosis but medial wedges increased plantar aponeurosis strain.

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 patellofemoral, tibiofemoral and Achilles tendon kinetics in female runners. Therefore, the aim of the current investigation was to examine the effects of orthoses with a 5° medial and lateral wedge on patellofemoral, Achilles tendon and KAM loading parameters during stance phase in female runners. A clinical investigation of this nature may provide insight into the potential efficacy of wedged foot orthoses for the prevention of knee and ankle pathologies in female runners.

Methods

Participants

Twelve healthy female recreational runners who trained at least 3 times/week over a minimum distance of 35 km (age 28.75 ± 6.69years, height 1.62 ± 0.06m and body mass 62.21 ± 3.31kg) volunteered to take part in this study. Each runner exhibited a rearfoot strike pattern as they exhibited an impact peak in their vertical ground reaction force curve. All identified as recreational runners, who trained a minimum of 3 times/week. Participants were also free from knee and pathology at the time of data collection and had not previously had any knee or ankle surgery. The participants provided written informed consent and the procedure was approved by a University, ethical panel (REF 357). The runners did not habitually utilize orthoses during their training activities.

Orthoses

Commercially available orthotics (Slimflex Simple, Algeos UK) made from Ethylene-vinyl acetate with a shore A rating of 65 were examined (Sinclair, 2018). The orthoses were modifiable, allowing either a 5° varus or valgus configuration spanning the full length of the device (Figure 1). To ensure consistency each participant wore the same footwear (Asics, Patriot 6) (Figure 2). The experimental footwear had a mean mass of 0.265 kg, heel thickness of 22 mm and heel drop of 10 mm. To prevent any order effects in the experimental data, participants ran in each orthotic condition in a counterbalanced manner. This was achieved by giving each orthotic condition a letter either A, B or C and presenting the orthoses in each of the six available sequences (ABC, ACB, BAC, BCA, CAB and CBA) to the first six participants, then repeating the process for the second six.

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Procedure

Participants ran over-ground at 3.5 m/s (Fukuchi et al., 2017), in three conditions (medial, lateral and no-orthotic), striking a piezoelectric force platform (Kistler, Kistler Instruments Ltd) sampling at 1000 Hz, with their right (dominant) foot. Limb dominance was determined by asking participants which foot that they would utilize to kick a ball. Running velocity was monitored using infrared timing gates (Newtest, Oy Finland) and a maximum deviation from the experimental running velocity was allowed. To ensure that a constant running velocity was measured with no evidence of targeting the force platform, the anterior-posterior ground reaction force was qualitatively examined following each trial, and the running trials were inspected visually for evidence of modification to the stride pattern. The stance phase was delineated as the duration over which >20 N vertical force was applied to the force platform. Runners completed five successful trials in each orthotic condition. Kinematic data was captured at 250 Hz via an eight camera motion capture system (Qualisys Medical AB, Goteburg, Sweden).

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 right foot, shank and thigh, retroreflective markers were placed unilaterally onto the 1st metatarsal, 5th metatarsal, calcaneus, medial and lateral malleoli, medial and lateral epicondyles of the femur. To define the pelvis segment, further markers were positioned onto the anterior (ASIS) and posterior (PSIS) superior iliac spines. The centers of the ankle and knee joints were delineated as the mid-point between the malleoli and femoral epicondyle markers (Graydon et al., 2015; Sinclair et al., 2015), whereas the hip joint centre was obtained using the positions of the ASIS markers (Sinclair et al., 2014). 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. To track the shank and thigh segments, carbon fiber tracking clusters comprising of four non-linear retroreflective markers were positioned onto these segments. Furthermore, the foot was tracked using the 1st metatarsal, 5th metatarsal and calcaneus markers and the pelvis using the ASIS and PSIS markers. Following marker placement, static calibration trials (not normalized to static trial posture) were obtained in each orthotic condition allowing for the anatomical markers to be referenced in relation to the tracking markers/ clusters.

Processing

Dynamic trials were digitized using Qualisys Track Manager then exported as C3D files to Visual 3D (C-Motion, Germantown, USA). Ground reaction force and kinematic data were smoothed using cut-off frequencies of 50 and 12Hz with a low-pass Butterworth 4th order zero-lag filter (Sinclair, 2018). Knee loading was examined through extraction of the peak KAM, peak patellofemoral contact force and contact stress, whereas ankle loading was explored by extracting the peak Achilles tendon force.

Patellofemoral force and stress were estimated using the model of Ward & Powers, (2004). This model has been shown to be sufficiently sensitive to resolve differences in patellofemoral loading between sexes (Sinclair & Selfe, 2015) and orthoses (Sinclair, 2018). Input parameters into the model were knee flexion angle, quadriceps moment arm, quadriceps force and knee extensor moment (Ho et al., 2012; van Eijden et al., 1986):

Firstly, an effective moment arm of the quadriceps muscle was quantified:

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

Quadriceps force was then estimated using the below formula:

$$\textbf{Quadriceps force} = \text{knee extensor moment} / \text{quadriceps moment arm}$$

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

$$\textbf{Patellofemoral contact force} = \text{quadriceps force} * \text{constant}$$

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)

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

Contact stress (MPa) was estimated as a function of the contact force divided by the sex specific patellofemoral contact areas as described by Besier et al., (2005):

$$\textbf{Patellofemoral contact stress} = \textbf{patellofemoral contact force} / \textbf{contact area}$$

Achilles tendon force was determined using the musculoskeletal model of Self and Paine (2001), which has been shown to be sufficiently sensitive to resolve differences in Achilles tendon loading between sexes (Greenhalgh & Sinclair, 2014) and orthoses (Sinclair et al., 2014). Input parameters into the model were ankle plantarflexion moment, ankle sagittal plane angle and Achilles tendon moment arm:

$$\textbf{Achilles tendon force} = \textbf{ankle plantarflexion moment} / \textbf{Achilles tendon moment arm}$$

$$\textbf{Achilles tendon moment arm} = \mathbf{-0.5910 + 0.08297} * \textbf{ankle sagittal plane angle} - \mathbf{0.0002606} * \textbf{ankle sagittal plane angle}^{\mathbf{2}}$$

Patellofemoral and Achilles tendon forces were normalized by dividing the net values by bodyweight (BW), whereas the KAM was normalized by dividing by body mass. Patellofemoral and Achilles tendon average load rate (BW/s) were quantified as the peak force divided by the time to peak force, whereas instantaneous load rate (BW/s) was determined as the maximum increase in force between frequency intervals. The KAM average load rate (Nm/kg/s) was quantified as the peak KAM divided by the time taken, whereas the instantaneous KAM load rate (Nm/kg/s) was determined the maximum increase between frequency intervals. The patellofemoral/ Achilles tendon (BW·s) and KAM (N/kg·s) impulse were calculated by multiplying the load during the stance phase by the stance phase duration.

Statistical Analyses

Means, standard deviations (SD) and 95 % confidence intervals (95% CI) were calculated for each outcome measurement for all three orthotic conditions. Differences between orthotic conditions were examined using one-way repeated measures ANOVA. Effect sizes were calculated using partial eta^2 (pη^2). Post-hoc pairwise comparisons were conducted on all significant main effects. In the event of a post-hoc comparison indicating statistical significance, the number of participants (N) who followed the direction of the statistical difference was reported. Finally, the mean difference and 95% CI of the difference between orthotic conditions for each outcome measurement were also calculated. Statistical actions were all conducted using SPSS v23.0 (SPSS, USA).

Results

Figure 3 and tables 1-2 present knee and ankle kinetic parameters as a function of different orthotic conditions.

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Achilles tendon kinetics

A main effect ($P<0.05$, $p\eta^2=0.27$) was evident for peak Achilles tendon force. Post-hoc analyses showed that peak Achilles tendon force was significantly larger in the no-orthotic condition ($P=0.03$, $N=9$) compared to the lateral orthosis (Figure 3a). A main effect ($P<0.05$, $p\eta^2=0.39$) was evident for Achilles tendon instantaneous load rate. Post-hoc analyses showed that Achilles tendon instantaneous load rate was significantly larger in the medial orthotic compared to the lateral ($P=0.003$, $N=11$) and no-orthotic ($P=0.03$, $N=10$) conditions. A main effect ($P<0.05$, $p\eta^2=0.30$) was shown for Achilles tendon impulse. Post-hoc analyses showed that Achilles tendon impulse was significantly larger in the no-orthotic ($P=0.004$, $N=11$) condition compared to the lateral orthosis.

Patellofemoral kinetics

A main effect ($P<0.05$, $p\eta^2=0.29$) was evident for the magnitude of peak patellofemoral force. Post-hoc pairwise comparisons showed that peak patellofemoral force was significantly larger in the medial condition compared to the lateral ($P=0.027$, $N=9$) and no-orthotic ($P=0.008$, $N=10$) conditions (Figure 3b). In addition, a main effect ($P<0.05$, $p\eta^2=0.26$) was found for peak patellofemoral stress. Post-hoc pairwise comparisons showed that peak patellofemoral force was significantly larger in the medial condition compared to the no-orthotic ($P=0.04$, $N=10$) condition (Figure 3c). A main effect ($P<0.05$, $p\eta^2=0.51$) was evident for the magnitude of patellofemoral instantaneous load rate. Post-hoc pairwise comparisons showed that patellofemoral instantaneous load rate was significantly larger in the medial ($P=0.00001$, $N=11$) and lateral ($P=0.03$, $N=9$) orthotic conditions compared to no-orthotic. Finally, a main effect ($P<0.05$, $p\eta^2=0.25$) was evident for the magnitude of patellofemoral impulse. Post-hoc pairwise comparisons showed that patellofemoral impulse was significantly larger in the medial orthotic in comparison to the lateral ($P=0.009$, $N=10$) orthotic conditions.

Knee kinetics

A main effect ($P<0.05$, $p\eta^2=0.28$) was evident for the magnitude of peak KAM. Post-hoc pairwise comparisons showed that peak KAM was significantly larger in the medial condition compared to the lateral ($P=0.03$, $N=9$) orthosis (Figure 3d). A main effect ($P<0.05$, $p\eta^2=0.39$) was also evident for the magnitude of KAM impulse. Post-hoc pairwise comparisons showed that KAM impulse was significantly larger in the medial ($P=0.001$, $N=9$) and no-orthotic ($P=0.02$, $N=11$) conditions in comparison to the lateral orthosis.

Discussion

The aim of the current investigation was to examine the effects of orthoses with a 5° medial and lateral wedge on knee and ankle joint kinetics in female runners. This represents the first investigation to compare the effects of medial/ laterally wedged orthoses on patellofemoral, Achilles tendon and KAM loading parameters in female runners.

The current study importantly demonstrated that peak patellofemoral stress was significantly greater when running with medial orthoses compared to the no-orthoses condition. This observation specifically supports the findings of Almonroeder et al., (2015) who observed increases in patellofemoral loading when running in medial orthoses. Similar to the suggestion presented by Almonroeder et al., (2015) and Sinclair, (2018), it is proposed that the increases in patellofemoral loading were mediated via an enhanced knee extension moment. The additional heel elevation provided by the orthotic conditions may have influenced the orientation of the ground reaction force vector such that the magnitude of the knee extensor moment, a key input parameter into the patellofemoral model was enhanced. This observation may be important regarding the initiation of patellofemoral pain, as the initiation of symptoms is mediated through excessive patellofemoral joint stress (Farrokhi et al., 2011). The findings from the current investigation indicate that running with medial orthoses may increase female runners' susceptibility to patellofemoral pain. This conclusion opposes those provided via previous randomized trials (Collins et al., 2008) and the recent meta-analytic review of Bonanno et al., (2017), which designate that foot orthoses are effective in preventing injuries. Therefore, further mechanistic trials are required to better understand the biomechanical causes responsible for the improvements in patellofemoral symptoms mediated via orthotic intervention.

In addition, peak KAM and the KAM impulse were significantly reduced in the lateral orthotic condition compared to the medial condition. This is in agreement with previous walking analyses described by Shimada et al., (2006); Hinman et al., (2008); Hinman et al., (2009); Jones et al., (2013), who also reported reductions in the KAM in when lateral orthoses were utilized. Furthermore, this observation agrees with the running observations of Lewinson et al., (2013) and Sinclair, (2018) who showed that laterally wedged orthoses significantly reduced the peak KAM. It is proposed that this observation is caused by the configuration of the lateral orthoses, which reduce the moment arm of the ground reaction force vector about the knee joint centre. An interesting qualitative observation is that of an early peak in the KAM waveform, which is present only when running in the medial and lateral orthoses (Figure 3d). It is proposed that this is a reflection of the increased stiffness of the orthoses, which are more firm than typical running shoe insoles (Janakiraman et al., 2011). This causes the rate at which the medial ground reaction force changes to increase, causing a discernible peak in the KAM curve in the orthotic conditions. Importantly the KAM is an effective measure of medial compartment loading (Birmingham et al., 2007), and both the peak KAM and KAM impulse are important predictors of knee OA (Miyazaki et al., 2002; Kean et al., 2012). Thus, it appears that the utilization of lateral orthoses may have potential to attenuate the risk of medial compartment knee OA in female runners.

The current investigation also revealed that Achilles tendon loading parameters were significantly reduced in the lateral orthotic condition. As lateral orthoses would be expected to increase the ankle eversion angle, this observation lends further weight to recent findings which oppose the long standing notion that hyper pronation augments the loads borne by the Achilles tendon. Indeed, in their prospective examination of 129 runners, Van Ginckel et al., (2009) found that lateral foot roll-over was a significant risk factor linked to the development of Achilles tendinopathy. As excessive tendon forces are the main stimulus for the initiation of Achilles tendinopathy (Abate et al., 2009), this finding may also have clinical relevance, and indicates that lateral orthoses may have the potential to be efficacious for female runners susceptible to Achilles tendinopathy.

A limitation in relation to the current investigation is that only the acute effects of wedged orthoses were examined in runners who did not habitually utilize foot orthoses. Therefore, although the lateral orthoses appear to attenuate tibiofemoral and Achilles tendon risk factors linked to the aetiology of chronic pathologies, it is currently unknown whether this will prevent or delay the initiation of injury symptoms. Furthermore, the duration over which the orthoses would need to be utilized in order to mediate a clinically meaningful change in patients is also not currently known. Although Hinman et al., (2008) found that the biomechanical effects of lateral orthoses do not appear to decline through continuous use, a longitudinal examination of these orthoses in runners would nonetheless be of practical and clinical relevance in the future. A further potential drawback is that it is only pain free controls were examined, meaning that only prophylactic inferences can be made in regards to the clinical efficacy of the orthoses examined in this study. Based on the observations of the current study it is important that forthcoming clinical investigations seek to examine the efficacy of lateral foot orthoses in runners with existing tibiofemoral and Achilles tendon pathologies. Future developments of this nature will help to determine the efficacy of wedged orthoses as treatment modalities for runners with chronic pathologies.

The current study adds to the current literature in the field of clinical biomechanics by providing a comprehensive examination of the effects of medial and lateral orthoses on knee and ankle loading parameters in female runners. The current investigation demonstrated that lateral orthoses reduced the magnitude of KAM and also the Achilles tendon force but that medial orthoses increased patellofemoral loading. The results from this study indicate that lateral orthoses may be effective in attenuating risk from medial tibiofemoral OA and Achilles tendinopathy, but medial wedge orthoses may increase the risk from patellofemoral pain in female runners.

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Competing interests

No conflict of interest will arise from any of the authors involved in this paper.

Author contributions

All named authors have made a significant and substantial contribution to all aspects of the study. Each of the named authors provided a meaningful contribution to the conception, design, execution and interpretation of the study data in addition to writing, drafting and revising the paper itself. This paper is submitted with the agreement and approval of both authors.

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

	Medial			**Lateral**			**No-orthotic**			
	Mean	SD	95% CI	Mean	SD	95% CI	Mean	SD	95% CI	
Peak Achilles tendon force (BW)	5.18	0.77	4.69-5.67	5.03 ***A***	0.71	4.58-5.47	5.34	0.85	4.80-5.88	*
Achilles tendon average load rate (BW/s)	44.87	10.28	38.47-51.40	43.39	8.48	38.00-48.78	46.83	12.60	38.82-54.83	
Achilles tendon instantaneous load rate (BW/s)	190.74 ***AB***	83.92	137.42-244.06	141.23	45.33	112.43-170.03	138.55	43.49	110.92-166.18	*
Achilles tendon impulse (BW·s)	0.55	0.10	0.48-0.62	0.52 ***A***	0.11	0.45-0.59	0.58	0.11	0.51-0.65	*
Peak patellofemoral force (BW)	2.77 ***AB***	0.74	2.30-3.24	2.62	0.80	2.11-3.12	2.60	0.76	2.11-3.08	*
Patellofemoral average load rate (BW/s)	23.85	7.32	19.21-28.50	22.25	7.03	17.78-26.71	22.53	7.71	17.63-27.43	
Patellofemoral instantaneous load rate (BW/s)	172.67 ***B***	63.59	132.27-213.08	159.14 ***A***	75.00	111.49-206.79	132.51	50.91	100.16-164.86	*
Patellofemoral impulse (BW·s)	0.22 ***B***	0.06	0.18-0.26	0.19	0.08	0.14-0.24	0.19	0.07	0.15-0.24	*
Peak patellofemoral stress (MPa)	7.37 ***A***	1.86	6.19-8.56	7.13	2.07	5.82-8.45	7.02	1.88	5.83-8.22	*
Peak KAM (Nm/kg)	1.14 ***B***	0.49	0.83-1.45	0.99	0.34	0.78-1.21	1.06	0.38	0.82-1.31	*
KAM average load rate (Nm/kg/s)	32.84	28.94	14.46-51.23	28.34	20.90	15.06-41.62	27.76	17.97	16.35-39.18	
KAM instantaneous load rate (Nm/kg/s)	100.61	60.05	62.46-138.77	93.36	45.87	64.21-122.50	94.20	45.50	65.29-123.11	
KAM impulse (Nm/kg·s)	0.08 ***B***	0.05	0.05-0.11	0.06 ***A***	0.04	0.04-0.09	0.08	0.04	0.06-0.11	*

Notes: * = significant main effect
A = significantly different from no-orthotic
B = significantly different from lateral orthotic

Table 2: Mean and 95% confidence interval differences between experimental conditions.

	Medial vs. Lateral		**Medial vs. No-orthotic**		**Lateral vs. No-orthotic**	
	Mean difference	95% CI difference	*Mean difference*	95% CI difference	*Mean difference*	95% CI difference
Peak Achilles tendon force (BW)	0.15	-0.16 – 0.46	-0.16	-0.34 – 0.03	***-0.31***	***-0.60 – -0.02***
Achilles tendon average load rate (BW/s)	1.48	-2.22 – 5.18	-1.95	-5.60 – 1.69	-3.43	-8.33 – 1.46
Achilles tendon instantaneous load rate (BW/s)	***49.52***	***21.03 – 78.00***	***52.19***	***5.58 – 98.80***	2.68	-21.73 – 27.08
Achilles tendon impulse (BW·s)	0.03	-0.02 – 0.09	-0.03	-0.07 – 0.01	***-0.06***	***-0.10 – -0.02***
Peak patellofemoral force (BW)	**0.15**	**0.05 – 0.26**	**0.17**	**0.02 – 0.31**	0.02	-0.16 – 0.19
Patellofemoral average load rate (BW/s)	1.61	-0.05 – 2.64	1.32	-0.32 – 2.97	-0.28	-1.51 – 0.95
Patellofemoral instantaneous load rate (BW/s)	13.53	-4.44 – 31.50	***40.16***	***25.10 – 55.23***	26.63	***3.59 – 49.67***
Patellofemoral impulse (BW·s)	***0.02***	***0.01 – 0.04***	0.02	-0.01 – 0.05	-0.001	-0.03 – 0.03
Peak patellofemoral stress (MPa)	0.24	-0.07 – 0.55	***0.35***	***0.01 – 0.69***	0.11	-0.36 – 0.58
Peak KAM (Nm/kg)	***0.15***	***0.003 – 0.29***	0.07	-0.10 – 0.25	-0.07	-0.18 – 0.03
KAM average load rate (Nm/kg/s)	4.50	-5.98 – 14.99	5.08	-9.50 – 19.65	0.58	-7.09 – 8.25
KAM instantaneous load rate (Nm/kg/s)	7.26	-16.71 – 31.23	6.41	-22.13 – 34.95	-0.85	-19.01 – 17.32
KAM impulse (Nm/kg·s)	***0.02***	***0.003 – 0.03***	-0.002	-0.02 – 0.01	-0.02	-0.03 – -0.008

Notes: Bold text = significant difference

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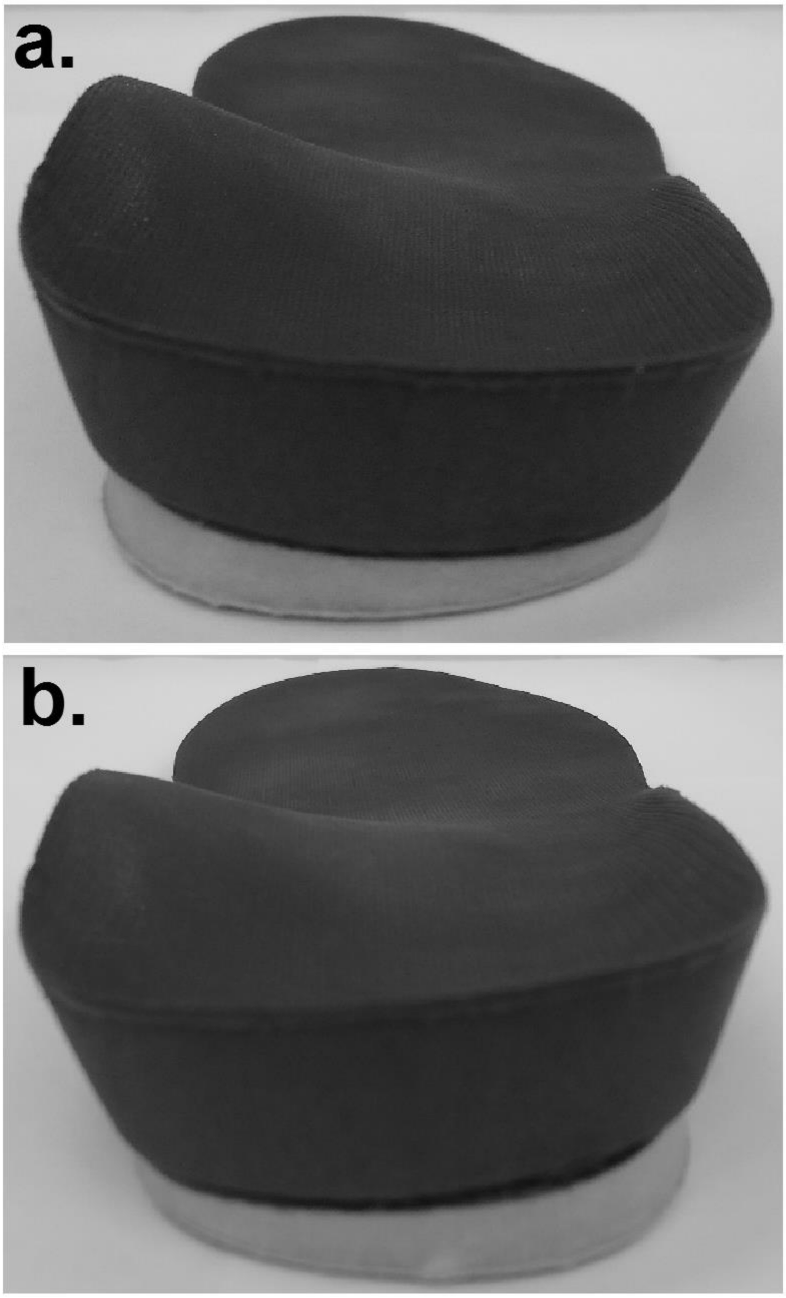

Figure 1: Experimental orthoses (a. = medial configuration and b. = lateral configuration).

Figure 2: Experimental footwear.

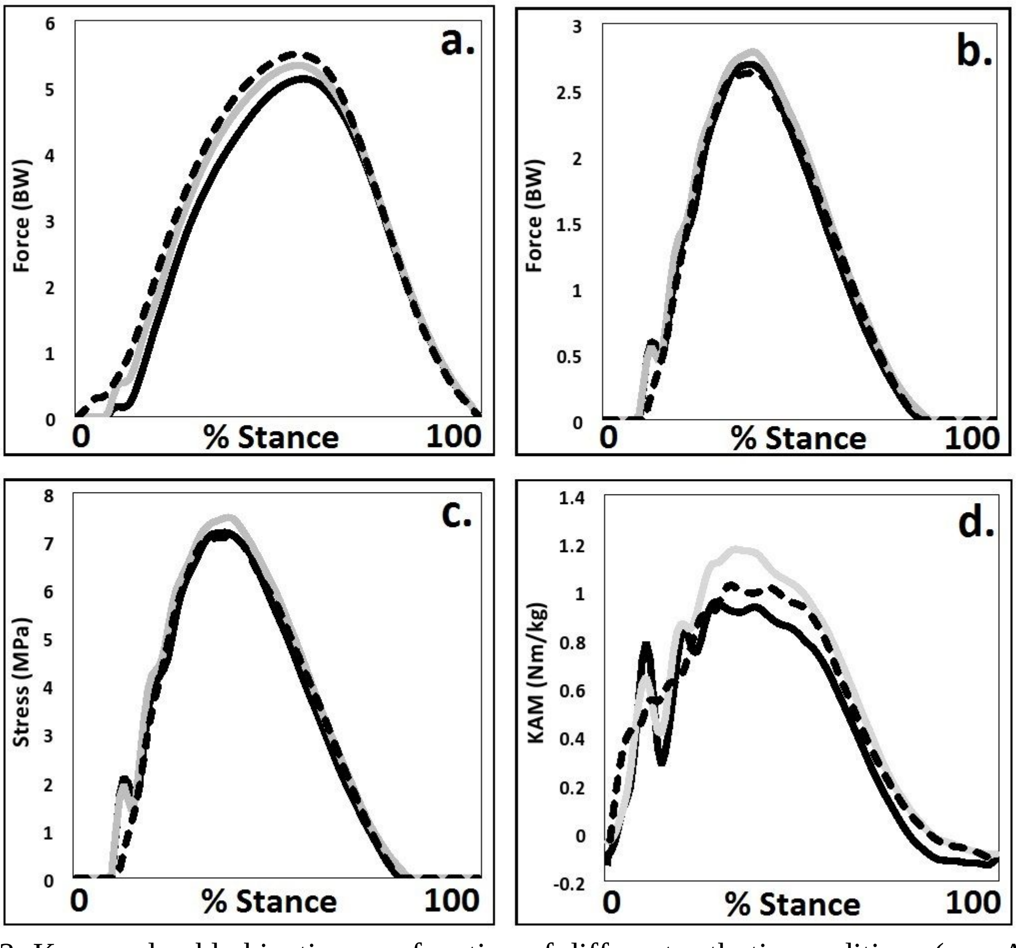

Figure 3: Knee and ankle kinetics as a function of different orthotic conditions (a. = Achilles tendon force, b. = patellofemoral force, c. = patellofemoral stress, d. = KAM) (Black = lateral, dot = no-orthotic, grey = medial).