Effects of toe-in/ out toe-in gait and lateral wedge orthoses on lower extremity joint kinetics; an exploration using musculoskeletal simulation and Bayesian contrasts.

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Abstract

INTRODUCTION: The aim of the current investigation was to examine the effects of both lateral orthoses and toe-in/ toe-out foot progression angles on lower extremity joint loading during walking using a musculoskeletal simulation approach.

METHODS: The current investigation examined 15 healthy males, walking in six different conditions (neutral, lateral orthoses, toe-in, lateral toe-in, toe-out and lateral toe-out). Walking kinematics were collected using an eight-camera motion capture system, and kinetics via an embedded piezoelectric force plate. Lower extremity joint loading was explored using a musculoskeletal simulation approach.

RESULTS: This investigation showed that peak patellofemoral joint stress was greater in the neutral (3.96 KPa/BW) and lateral orthoses (4.20 KPa/BW) conditions compared to toe-in (3.33 KPa/BW), lateral toe-in (3.43 KPa/BW), toe-out (3.35 KPa/BW) and lateral toe-out (3.53 KPa/BW) and ankle joint impulse larger in the toe-in (1.65BW·s) and toe-out (1.62BW·s) foot progression angle modalities compared to neutral (1.51BW·s) and lateral orthoses (1.53BW·s). Furthermore, it was also shown that medial tibiofemoral impulse was statistically greater in the toe-in (1.20BW·s) and lateral toe-in (1.15BW·s) conditions compared to neutral (1.07BW·s), lateral orthoses (1.07BW·s), toe-out (1.09BW·s) and lateral toe-out (1.05BW·s).

CONCLUSIONS: Therefore, the current investigation provides evidence that altering the foot progression angle may attenuate the risk from patellofemoral disorders whilst simultaneously enhancing the risk from degenerative ankle pathologies. Similarly, adopting a toe-in foot progression angle may also increase the risk from medial tibiofemoral degeneration.

Introduction

Walking is undoubtably a fundamental aspect of everyday living, and the primary locomotion modality in humans. The knee joint plays an important role in load bearing during walking and other common daily activities (1). However, excessive knee joint forces are regarded as the contributing factor to the initiation and progression of degenerative knee disorders (2-3). Therefore, due to factors such as age, excessive body mass or previous injury, the negative effects of excessive knee joint loads begin to emerge and the risk of knee disorders such as osteoarthritis (OA) increase (4).

Knee OA represents a degenerative articular disease, caused by erosion and deterioration of the articular cartilage within the knee joint itself (5), and those with knee OA experience ongoing pain and stiffness (6). OA at the knee joint has been shown to be present in as many as 10% of individuals over the age of 55, and importantly over 90% of knee OA cases are observed in the medial tibiofemoral compartment (7-8). This is because during traditional activities of daily life over 60% of the total load borne by the tibiofemoral joint passes through the medial compartment of the knee (9).

Owing to the proposed association between excessive joint forces and the initiation/ progression of joint degeneration, as well as the high incidence of medial tibiofemoral OA, several strategies have emerged that aim to lower medial tibiofemoral joint loading and in turn the risk from OA. Because of the challenges associated with the quantification of tibiofemoral contact forces, the knee adduction moment is typically adopted as a proxy for medial tibiofemoral loading (10-11). A commonly adopted strategy is the utilization of lateral wedge insoles/ orthoses (12), whereby the centre of pressure is forced into a more lateral position; causing a reduction in the lever arm of the knee adduction moment (13). Importantly lateral wedge insoles have been shown to attenuate the magnitude of the knee adduction moment during gait (14-15), stair ascent and descent (16).

A further conservative modality for the reduction of the knee adduction moment is alterations to individuals walking gait pattern (1). Changing the foot progression angle has two variants; toe-out and toe-in gait, which unlike other gait modification techniques appear to be sustainable up to 10 weeks (17). Toe-in (represented by internal rotation of the foot) and toe-out (characterized by external rotation of the foot) gait patterns are similarly designed to influence the knee adduction moment by moving the centre of pressure mediolaterally (17). Typically toe-in gait patterns have been shown to reduce the magnitude of the first peak in the knee adduction moment time curve (1; 18-19), yet toe-out gait has correspondingly been found to reduce the second knee adduction moment peak (1; 18-20). However, conversely Jenkyn et al., (2008) showed that that toe-out walking reduced both the first and second peak of the knee adduction moment curve. Similarly, both Van den Noort et al., (18) and Khan et al., (1) showed that the knee adduction moment impulse was reduced with a toe-in gait pattern, yet other investigations have shown that there is no effect of altering the foot progression angle on this parameter (17).

As the knee adduction moment is a pseudo measurement for tibiofemoral loading, all of the previous analyses concerning the effects of lateral orthoses and gait modifications have used this measurement. Although the knee adduction moment and its derivatives have been linked to medial knee joint cartilage degeneration (10), joint moments are not representative of localized joint contact loads (21). Importantly a recent investigation using instrumented knee prostheses showed only moderate correlations between the knee adduction moment and direct medial tibiofemoral joint loading magnitudes during walking (3). Indeed Herzog et al., (21) identified importantly that muscle forces are the primary contributors to joint forces and in recent years advances in musculoskeletal simulation software have allowed allow muscle driven calculations of lower extremity joint reaction forces (22). However, such approaches have not yet been utilized to explore the effects of lateral orthoses and foot progression angle gait modifications.

Similarly, whilst the effects of both lateral orthoses and modified foot progression angle gait patterns have been examined previously, they have focussed only on indices of medial tibiofemoral joint loading (quantified using the knee adduction moment and its derivatives). Both wedged foot orthoses and alterations in the foot progression angles during gait are likely to mediate alterations at more than one body segment and thus more than one joint (23). Therefore, potential positive alterations in joint loading mediated at the medial compartment of the tibiofemoral joint may concurrently negatively impact other lower extremity joints.

To summarize, there is currently no scientific research that has explored the effects of lateral orthoses/ foot progression angle modifications on collective lower extremity joint loading using a musculoskeletal simulation-based approach. Therefore, the aim of the current investigation was to examine the effects both lateral orthoses and foot progression angles on lower extremity joint loading during walking using a musculoskeletal simulation approach. A study of this nature may provide further insight into the cumulative biomechanical effects of different modalities designed to reduce the risk from medial tibiofemoral OA.

The current investigation tests the hypothesis firstly that lateral orthoses and a toe-in gait pattern will reduce medial tibiofemoral loading and secondly that these parameters when used collectively will serve to further attenuate medial tibiofemoral forces during walking.

Methods

Participants

Fifteen male participants (age 31.73±4.96 years, height 1.72±0.06 m and body mass 69.31±9.92 kg and BMI = 23.45±2.78 kg/m^2). volunteered to take part in this study. All participants were free from pathology at the time of data collection and provided written informed consent, in accordance with the principles outlined in the Declaration of Helsinki. The inclusion criteria for the subjects included healthy adults, aged 20–45 years and having a BMI of less than 30 kg/m^2 (1). The participants were excluded on the basis of any musculoskeletal disorder, previous knee surgery or inability to adopt the novel gait pattern. The procedure utilized for this investigation was approved; by a university ethical committee (STEMH 1013).

Experimental orthoses

Commercially available full-length orthoses (Slimflex Simple, High Density, Full Length, Algeos UK) were examined in the current investigation. 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 featured a 5° lateral wedge configuration which spanned the full length of the device. To ensure consistency each participant wore the same footwear (Asics, Gel Patriot 6). The experimental footwear had a mean mass of 0.265 kg, heel thickness of 22 mm and heel drop of 10 mm.

Foot progression angles

As this study focused on the effects of toe-in and toe-out foot progression angles, values of 15° during the stance phase were targeted (1). Therefore, the force plate was marked with a straight line that ran directly through the middle of the long axis of the force plate (neutral), a 15° line that represented the foot progression angle required for the toe-out condition for the right foot (toe-out) and a further 15° line that represented the foot progression angle required for the toe-in condition for the right foot (toe-in). Participants were given a 5-minute period of familiarization for each of the three aforementioned settings in order to become accustomed to the experimental conditions (1). Following this the force plate markings were removed prior to the commencement of data collection and participants were asked to replicate these foot progression angles during data collection. This process was deemed to be ecologically valid as individuals seeking to implement these modifications into their own gait mechanics would not have lines drawn or feedback for each footfall during normal walking (1).

To quantify the foot progression angle during data collection, the path of the centre of pressure was determined from using the force plate. The progression angle of the foot was calculated by intersecting the position of the centre of pressure at the time of foot contact with the centre of pressure at toe-off. The toe-out (represented by a positive value) and toe-in angles (represented by a negative value) were determined as the angle formed by the intersection line relative to the directly anterior direction (24).

Procedures

Participants walked without orthoses in three conditions; straight foot (neutral), a toe-out foot progression angle (toe-out) and toe-in foot progression angle (toe-in) and also with orthoses in the same conditions; lateral orthoses, lateral toe-in and lateral toe-in. Participants walked at a velocity of 1.5 m/s (±5%), striking an embedded piezoelectric force platform (Kistler, Kistler Instruments Ltd) with their right (dominant) foot. Walking velocity was monitored using infrared timing gates (Newtest, Oy Koulukatu). The stance phase was delineated as the duration over which 20 N or greater of vertical force was applied to the force platform (25). Participants completed five successful trials in each condition and the order that participants walked in each footwear condition was counterbalanced. Kinematics and ground reaction forces data were synchronously collected. Kinematic data was captured at 250 Hz via an eight-camera motion analysis system (Qualisys Medical AB) and ground reaction forces captured at 1000 Hz. Dynamic calibration of the motion capture system was performed before each data collection session.

To define the anatomical frames of the thorax, pelvis, thighs, shanks and feet retroreflective markers were placed at the C7, T12 and xiphoid process landmarks and also positioned bilaterally onto the acromion process, iliac crest, anterior superior iliac spine (ASIS), posterior super iliac spine (PSIS), medial and lateral malleoli, medial and lateral femoral epicondyles, greater trochanter, calcaneus, first metatarsal and fifth metatarsal. Carbon-fiber tracking clusters comprising of four non-linear retroreflective markers were positioned onto the thigh and shank segments. In addition to these the foot segments were tracked via the calcaneus, first metatarsal and fifth metatarsal, the pelvic segment was tracked using the PSIS and ASIS markers and the thorax segment was tracked using the T12, C7 and xiphoid markers.

Static calibration trials were obtained with the participant in the anatomical position in order for the positions of the anatomical markers to be referenced in relation to the tracking clusters/markers. A static trial was conducted with the participant in the anatomical position in order for the anatomical positions to be referenced in relation to the tracking markers, following which those not required for dynamic data were removed.

Processing

Dynamic trials were digitized using Qualisys Track Manager, in order to identify anatomical and tracking markers and then exported as C3D files to Visual 3D (C-Motion, Germantown, MD). All data were normalized to 100 % of the stance phase. Ground reaction force (GRF) and kinematic data were smoothed using cut-off frequencies of 25 and 6 Hz with a low-pass Butterworth 4th order zero lag filter (26).

Data during the stance phase were exported from Visual 3D into OpenSim 3.3 software (Simtk.org). A validated musculoskeletal model with 12 segments, 19 degrees of freedom and 92 musculotendon actuators (27) was used to estimate lower extremity joint forces. The model featured the same segments and muscle tendon units as the standard Gait2392 Opensim model (***https://simtk-confluence.stanford.edu:8443/display/OpenSim/Gait+2392+and+2354+Models?preview=/3376103/3736767/Gait%202392%20vs.%20Gait%202354.pdf***) but the tibiofemoral joint was separated into medial and lateral compartments to allow joint reaction analyses at these areas separately and the model also featured a patella. The model was firstly scaled for each participant to account for the anthropometrics of each individual. Then as muscle forces are the main determinant of joint compressive forces (21), muscle kinetics were quantified using static optimization. The static optimization simulation process calculates the muscle forces required in order to recreate the experimentally measured motions. Compressive ankle, medial/ lateral tibiofemoral, patellofemoral and hip joint forces were calculated via the joint reaction analyses function within OpenSim using the muscle forces generated from the static optimization process as inputs. The joint reaction analysis function in OpenSim calculates the joint loads transferred between two contacting bodies, about the joint centre location identified during the static trial (28). In the current investigation, joint forces were normalized by dividing by each participants body weight (BW). Compressive hip joint forces were representative of the contact forces between the femur and acetabular cartilage, tibiofemoral forces between the medial/ lateral tibial and femoral cartilage, patellofemoral joint forces between the femur and patella cartilage and ankle joint forces between the tibia and talar cartilage. Patellofemoral joint stress (KPa/BW) was also quantified by dividing the patellofemoral joint reaction force by the patellofemoral contact area. Patellofemoral contact areas were obtained by fitting a polynomial curve to the sex specific data of Besier et al., (29). From the above processing, peak normalized ankle force, peak lateral tibiofemoral force, and peak patellofemoral force/ stress during the stance phase were extracted for statistical analyses. In addition, as the hip and medial tibiofemoral joint force curves exhibit a double peaked pattern, the normalized values at the first and second peak for these parameters were extracted for analysis.

In addition, ankle, medial/ lateral tibiofemoral, patellofemoral and hip instantaneous load rates (BW/s) were also extracted by obtaining the peak increase in force between adjacent data points. Finally, ankle, medial/ lateral tibiofemoral, patellofemoral and hip force impulses (BW·s) and stresses (KPa/BW·s) during the stance phase were calculated using a trapezoidal function.

Finally, in order to determine the mechanisms responsible for any alterations in joint loading, the forces (BW) for the major muscles crossing the hip, knee and ankle joints were quantified at the instances of the aforementioned peak joint forces/ stresses. Similarly, for any joint whereby only statistical alterations in the joint impulse were observed between conditions, the muscle force impulses for the major muscles crossing the associated joint were also extracted.

Statistical analyses

Following data processing, compressive joint forces (hip, patellofemoral, ankle, medial/ lateral tibiofemoral), during the stance phase were exported and temporally normalized using linear interpolation to 101 data points for statistical analysis using Statistical parametric mapping (SPM). In agreement with Pataky et al. (30), SPM was implemented in a hierarchical manner, analogous to one-way repeated measures ANOVA with post-hoc t-tests. Therefore, the entire data-set was examined first, and if a statistical main effect was reached then post-hoc tests comparing individual conditions were conducted on each component separately. For discrete parameters (peak joint forces, joint impulse, joint instantaneous load rates, muscle forces and muscle force impulses), means and standard deviations were calculated for each condition. Differences in discrete biomechanical parameters were examined using Bayesian one-way repeated measures ANOVA with default prior scales using JASP software 0.10.2 (31). In the event of a main effect, post-hoc Bayesian paired t-tests were conducted between each condition. Similarly, relationships between 1. discrete joint loads and foot progression angle and 2. discrete joint loads and muscle forces at the instances of peak joint loads/ muscle force impulses across all conditions were examined using Bayesian Pearson's correlation analyses using SPSS 27.0 software (SPSS, IBM). Bayesian factors (BF) were used to explore the extent to which the data supported the alternative (H_1) hypothesis. Bayes factors throughout were interpreted in accordance with the recommendations of Jeffreys, (32), with values ≥3 indicating sufficient evidence in support of H_1. In the interests of conciseness only variables that present with Bayes factors ≥3 and SPM analyses showing statistical significance will be presented.

Results

<u>*Foot progression angles*</u>

The mean ± SD of foot progression angles were: neutral = 0.68 ± 2.70°, lateral orthoses = 1.47 ± 2.85°, toe-out = 12.38 ± 4.42°, lateral toe-out = 12.44 ± 3.58°, toe-in = -10.63 ± 2.82° and lateral toe-in = -10.54 ± 2.37°.

<u>*Discrete analyses – joint loading*</u>

<u>***@@@Table 1 near here@@@***</u>

For the first peak of the hip force there was decisive evidence of a main effect (BF = 129890.84). Post-hoc analyses showed that values were larger in the lateral orthoses (BF = 42.84), neutral (BF = 35.68), toe-out (BF = 22.47), lateral toe-out (BF = 10.91) conditions compared to toe-in. Furthermore, values were also larger in the lateral orthoses (BF = 149.13), neutral (BF = 7.06), toe-out (BF = 3.67), lateral toe-out (BF = 22.12) conditions compared to lateral toe-in (Table 1). For the second peak of the hip force there was decisive evidence of a main effect (BF = 24.93). Post-hoc analyses showed that values were larger in the lateral orthoses (BF = 9.28), neutral (BF = 7.37), toe-in (BF = 5.93) and lateral toe-in (BF = 3.04) conditions compared to toe-out. Furthermore, values were also larger in the lateral orthoses (BF = 16.21), neutral (BF = 32.86) and lateral toe-out (BF = 3.16) conditions compared to lateral toe-out (Table 1).

For the first peak of the medial tibiofemoral force there was decisive evidence of a main effect (BF = 298.86). Post-hoc analyses showed that values were larger in the neutral condition compared to toe-in (BF = 5.82), lateral toe-in (BF = 4.73), toe-out (BF = 229.47) and lateral toe-out (BF = 20.24). In addition, values for this parameter were larger in lateral orthoses compared to lateral toe-in (BF = 4.08), toe-out (BF = 14.12) and lateral toe-out (BF = 3.54) (Table 1). For the second peak of the medial tibiofemoral force there was also decisive evidence of a main effect (BF = 8543.01). Post-hoc analyses showed that values were larger in the toe-in condition compared to lateral orthoses (BF = 248.62), neutral (BF = 4.30), toe-out (BF = 5.27) and lateral toe-out (BF = 15.67). In addition, values for this parameter were larger in the lateral toe-in condition compared to lateral orthoses (BF = 77.03) and lateral toe-out (BF = 3.27) (Table 1). For the medial tibiofemoral force impulse there was decisive evidence of a main effect (BF = 499513.34). Post-hoc analyses showed that values were larger in the toe-in condition compared to lateral orthoses (BF = 14.72), neutral (BF = 13.71), toe-out (BF =

175.33) and lateral toe-out (BF = 783.24). In addition, values for this parameter were larger in the lateral toe-in condition compared to lateral orthoses (BF = 621.47) and lateral toe-out (BF = 236.42) (Table 1).

For the lateral tibiofemoral force impulse there was decisive evidence of a main effect (BF = 1442262604595.22). Post-hoc analyses showed that values were larger in the lateral orthoses condition compared to toe-in (BF = 225.77) and lateral toe-in (BF = 66.20) and in the neutral compared to toe-in (BF = 31.14) and lateral toe-in (BF = 25.76). In addition, values for this parameter were larger in the toe-out condition compared to lateral orthoses (BF = 4.63), neutral (BF = 7.74), toe-in (BF = 10906.26) and lateral toe-in (133.70) and also in the lateral toe-out compared to lateral orthoses (BF = 11.82), neutral (BF = 34.14), toe-in (BF = 2393.76) and lateral toe-in (1151.05) (Table 1).

For the peak patellofemoral force there was decisive evidence of a main effect (BF = 132.68). Post-hoc analyses showed that values were larger in the neutral condition compared to toe-in (BF = 5.82), lateral toe-in (BF = 4.73), toe-out (BF = 229.47) and lateral toe-out (BF = 20.24). In addition, values for this parameter were larger in lateral orthoses compared to lateral toe-in (BF = 4.08), toe-out (BF = 14.12) and lateral toe-out (BF = 3.54) (Table 1).

For the peak patellofemoral stress there was decisive evidence of a main effect (BF = 3868.64). Post-hoc analyses showed that values were larger in the neutral condition compared to toe-in (BF = 11.90), lateral toe-in (BF = 65.92), toe-out (BF = 93.58) and lateral toe-out (BF = 83.13). In addition, values for this parameter were larger in lateral orthoses compared to toe-out (BF =

19.19) (Table 1). In addition, for the patellofemoral stress impulse there was substantial evidence of a main effect (BF = 3.87). Post-hoc analyses showed that values were larger in the neutral condition compared to toe-out (BF = 31.48) and lateral toe-out (BF = 212.27). In addition, values for this parameter were larger in lateral orthoses compared to toe-out (BF = 4.88) and lateral toe-out (BF = 29.24) (Table 1).

For the ankle force impulse there was decisive evidence of a main effect (BF = 12448.22). Post-hoc analyses showed that values were larger in the toe-out condition compared to neutral (BF = 8.43) and lateral orthoses (BF = 152.30). In addition, values for this parameter were larger in the toe-in condition compared to neutral (BF = 52.45) and lateral orthoses (BF = 184.65) and also in the lateral toe-in condition compared to lateral orthoses (BF = 34.73) (Table 1).

Discrete analyses – muscles forces

@@@Table 2 near here@@@

For the gluteus medius 2 at the instance of the first peak of the hip force there was decisive evidence of a main effect (BF = 381404.40). Post-hoc analyses showed that values were larger in the lateral orthoses (BF = 10.46), neutral (BF = 755.02), toe-out (BF = 10.78), lateral toe-out (BF = 57.35) conditions compared to toe-in. Furthermore, values were also larger in the lateral orthoses (BF = 87.33), neutral (BF = 19.43), toe-out (BF = 98.26), lateral toe-out (BF = 251.58) conditions compared to lateral toe-in (Table 2). Similarly, for the gluteus medius 3 at the instance of the first peak of the hip force there was decisive evidence of a main effect (BF = 732.05). Post-hoc analyses showed that values were larger in the lateral orthoses (BF = 103.36), neutral (BF = 10.60), toe-out (BF = 28.17), lateral toe-out (BF = 124.47) conditions compared to lateral toe-in (Table 2). For the gluteus minimus 3 at the instance of the first peak of the hip force there was substantial evidence of a main effect (BF = 6.02). Post-hoc analyses showed that values were larger in the lateral orthoses (BF = 46.62), neutral (BF = 10.02), toe-out (BF = 4.89), lateral toe-out (BF = 6.48) conditions compared to lateral toe-in (Table 2). For the tensor fasciae latae at the instance of the first peak of the hip force there was decisive evidence of a main effect (BF = 14450.22). Post-hoc analyses showed that values were larger in the lateral orthoses (BF = 30.00), neutral (BF = 22.22), toe-out (BF = 4102.04), lateral toe-out (BF = 47.70) conditions compared to lateral toe-in. Furthermore, values were also larger in the toe-out compared to lateral orthoses (BF = 14.80) and toe-in (BF = 48.83) conditions (Table 2). For the rectus femoris at the instance of the first peak of the hip force there was decisive evidence of a main effect (BF = 721.75). Post-hoc analyses showed that values were larger in the lateral orthoses (BF = 14.97), toe-out (BF = 777619.80) and lateral toe-out (BF = 12.80) conditions compared to lateral toe-in. Furthermore, values were also larger in the toe-out compared to lateral orthoses (BF = 16.12) and toe-in (BF = 518.30) conditions (Table 2).

For the rectus femoris at the instance of the second peak of the hip force there was substantial evidence of a main effect (BF = 3.56). Post-hoc analyses showed that values were larger in the toe-in condition compared to toe-out (BF = 8.78) and in the lateral orthoses (BF = 4.03) and toe-in (BF = 6.41) conditions compared to lateral toe-out (Table 2).

For the vastus intermedius at the instance of the first peak of the medial tibiofemoral force, there was substantial evidence of a main effect (BF = 5.72). Post-hoc analyses showed that values were larger in the lateral orthoses (BF = 82.13) and neutral (BF = 5.76) conditions compared to toe-out (Table 2). For the vastus lateralis at the instance of the first peak of the medial tibiofemoral force, there was substantial evidence of a main effect (BF = 8.22). Post-hoc analyses showed that values were greater in the lateral orthoses (BF = 25.35) and neutral (BF = 6.58) conditions compared to toe-out and also in the neutral compared to toe-in (BF = 3.16) and lateral toe-out (BF = 4.92) (Table 2). For the vastus medialis at the instance of the first peak of the medial tibiofemoral force, there was substantial evidence of a main effect (BF = 4.29). Post-hoc analyses showed that values were larger in the lateral orthoses (BF = 186.62) and neutral (BF = 5.64) conditions compared to toe-out (Table 2). For the rectus femoris at the instance of the first peak of the medial tibiofemoral force, there was strong evidence of a main effect (BF = 14.77). Post-hoc analyses showed that values were larger in the lateral orthoses (BF = 58.18), toe-in (BF = 109.18) and lateral toe-in (BF = 79.75) conditions compared to toe-out and also in the toe-in (BF = 7.40) compared to lateral toe-out condition (Table 2).

For the lateral gastrocnemius at the instance of the second peak of the medial tibiofemoral force, there was decisive evidence of a main effect (BF = 1023699.85). Post-hoc analyses showed that values were larger in the lateral orthoses (BF = 59.96), toe-in (BF = 30.04) and lateral toe-in (BF = 17.11) conditions compared to toe-out. Similarly, values were also larger in the lateral orthoses (BF = 1601.74), toe-in (BF = 240.40) and lateral toe-in (BF = 44.13) compared to lateral toe-out (Table 2).

For the vastus intermedius at the instance of peak patellofemoral stress, there was substantial evidence of a main effect (BF = 3.61). Post-hoc analyses showed that values were greater in the lateral orthoses (BF = 22.61) and neutral (BF = 6.85) conditions compared to toe-out and also in the neutral (BF = 3.90) compared to lateral toe-out (Table 2).

For the lateral gastrocnemius impulse there was decisive evidence of a main effect (BF = 36188.95). Post-hoc analyses showed that values were greater in the toe-out compared to lateral orthoses (BF = 276.36), neutral (BF = 24.91) and toe-in (BF = 38.24) also in the lateral toe-out compared to lateral orthoses (BF = 80189.72), neutral (BF = 33.57) and toe-in (BF = 154.09) conditions. There was also very strong evidence of a main effect for medial gastrocnemius impulse (BF = 37.47). Post-hoc analyses showed that values were greater in the toe-out compared to lateral orthoses (BF = 18.87) and neutral (BF 5.22) and also in the lateral toe-out compared to lateral (BF = 99.06) and neutral (BF = 35.27) conditions. There was very strong evidence of a main effect for tibialis anterior impulse (BF = 49.40). Post-hoc analyses showed that values were larger in the toe-out condition compared to toe-in (BF = 17.27) and lateral toe-in (BF 502.53). Finally, there was also decisive evidence of a main effect for tibialis posterior impulse (BF = 645123.12). Post-hoc analyses showed that values were greater in the toe-out condition compared to lateral orthoses (BF = 66.82), neutral (BF = 8.69) toe-in (BF = 656.32) and lateral toe-in (BF = 625.85) and also in the lateral orthoses (BF = 64.81), neutral (BF = 12.78) toe-in (BF = 580.43) and lateral toe-in (BF = 829.33) conditions.

Statistical parametric mapping

The analysis of the overall data set using SPM revealed significant main effects for hip force, medial tibiofemoral force, lateral tibiofemoral force, patellofemoral force and patellofemoral stress thus post-hoc investigation between individual conditions was required (Supplemental figure 1).

For hip force, lateral orthoses were associated with increased hip joint loading from during early stance compared to both toe-in and lateral toe-in, but reduced loading in late stance in relation to lateral toe-in (Figure 1). Furthermore, the neutral condition was associated with increased hip loading during early stance compared to toe-in, lateral toe-in and lateral toe-out and in late stance compared to toe-out and lateral toe-out (Figure 1). In addition, the toe-in condition exhibited increased hip loading during late stance but reduced loading during early stance compared to toe-out and lateral toe-out (Figure 1). In addition, the lateral toe-in condition exhibited increased hip loading during late stance but reduced loading during early-mid stance (Figure 1).

For patellofemoral force, lateral orthoses were associated with increased patellofemoral loading during early and late stance compared to toe-out (Figure 2). Similarly, the neutral condition exhibited increased patellofemoral loading during early stance compared to toe-out and during early and late stance compared to lateral toe-out (Figure 2). Furthermore, for patellofemoral stress lateral orthoses were associated with increased stress during early stance compared to toe-in and toe-out and during early and late stance in relation to lateral toe-out (Figure 2).

For medial tibiofemoral force, lateral orthoses were associated with increased medial loading during early stance compared to toe-in and toe-out and reduced loading during late stance compared to toe-in and lateral toe-out (Figure 3). In addition, the neutral condition was associated with increased loading during early stance yet reduced loading in late stance compared to toe-in, lateral toe-in, toe-out and lateral toe-out conditions (Figure 3). Furthermore, the toe-in and lateral toe-in conditions were associated with increased loading during late stance but decreased loading in early stance (Figure 3).

For lateral tibiofemoral force, lateral orthoses were associated with increased lateral loading during early-late stance compared to toe-in and lateral toe-in (Figure 4ab). In addition, lateral orthoses were associated with increased loading during early stance compared to lateral toe-out yet reduced loading during mid-late stance compared to toe-out and lateral toe-out (Figure 4cd). Similarly, the neutral condition was associated with increased lateral loading during early-late stance compared to toe-in and lateral toe-in (Figure 4). In addition, the neutral condition was associated with increased loading during early stance compared to lateral toe-out yet reduced loading during late stance compared to toe-out and lateral toe-out (Figure 4). Furthermore, the toe-in condition and lateral toe-in conditions were associated with increased loading in early stance but enhanced loading during late stance compared to toe-out and lateral toe-out (Figure 4).

Correlation analyses

@@@Table 3 near here@@@

@@@Table 4 near here@@@

Correlations between the foot progression angle and joint loading indices and the associated Bayes factors are presented in Table 3. Similarly, correlations between joint forces/ impulses and muscle forces/ impulses at the instances of peak joint loads/ impulses and the associated Bayes factors are presented in Table 4.

Discussion

The aim of the current investigation was to examine the effects both lateral orthoses and foot progression angles on lower extremity joint loading during walking using a musculoskeletal simulation approach. As the first investigation to examine the cumulative effects of lateral orthoses/ foot progression angles on lower extremity joint loading a study of this nature may provide further insight into the biomechanical effects of different modalities designed to reduce the risk from medial tibiofemoral OA.

At the medial aspect of the tibiofemoral joint, the findings do not support the hypothesis and similarly oppose those of Jones et al., (14) and Hinman et al., (15) in that lateral orthoses did not statistically influence joint loading at the medial tibiofemoral joint in relation to no-orthoses conditions. However, it is important to recognise that both Jones et al., (14) and Hinman et al., (15) examined participants with existing medial tibiofemoral OA, so biomechanical responses to lateral orthoses may be condition dependent. Furthermore, both the discrete and SPM based analyses showed a contradictory pattern of statistical differences in relation to the influence of the foot progression angle. Specifically, in partial agreement with previous findings, the first medial tibiofemoral force peak was greater in the neutral and lateral orthoses conditions whereas the second peak was statistically larger in the toe-in conditions (1; 18-20). This observation concurs with the correlation analysis between foot progression angle and the second medial tibiofemoral force peak and also the correlational analyses between joint and muscle kinetics (24). Therefore, as the first peak was best associated with the vasti muscle kinetics, the reductions in vasti forces at the first peak in the toe-in and toe-out conditions provides clear insight into the mechanisms responsible for the reductions in the first medial tibiofemoral force peak in these conditions (21). However, changes in the second medial tibiofemoral peak cannot be explained via the correlational analyses between joint/ muscle kinetics.

Importantly, in addition to differences at each of the force peaks, in opposition to our hypothesis, this study showed a main effect for the medial tibiofemoral impulse across the stance phase, which was found to be greater in the toe-in conditions. However, both SPM and discrete analyses also showed that lateral tibiofemoral loading was statistically attenuated in the toe-in conditions. This observation similarly is supported by the correlation analyses between foot progression angle and the peak lateral tibiofemoral force and also the medial tibiofemoral force impulse, and indicates that reductions in joint loading borne by the medial compartment during the stance are simply shifted to the lateral aspect of the joint. However, given the greatly enhanced incidence of medial tibiofemoral OA (7) and the association between knee joint loading and initiation of cartilage degeneration (3), the current investigation indicates that toe-in gait patterns increase risk from the mechanisms linked to the aetiology of medial knee OA.

Much like at the medial tibiofemoral compartment, both the discrete and SPM based analyses at the hip joint showed a contradictory pattern of statistical differences. Firstly, in the neutral and lateral orthoses conditions as well as the toe-out and lateral toe-out conditions, the smaller first peak of the hip joint contact force was enhanced compared to the toe-in and lateral toe-in conditions. Conversely, it was also revealed that neutral and lateral orthoses conditions as well as the toe-in and lateral toe-in modalities were associated with increases in the larger second peak of the hip joint contact force in relation to the toe-out conditions. Importantly muscle forces are the primary determinant of joint loading (21). Therefore the observations from the correlational analyses showing that the first peak was best associated with adductor magnus and gluteal muscle forces and the second peak with adductor magnus, gluteal, rectus femoris and tensor fasciae latae provide insight into the mechanisms responsible for the aforementioned changes in hip joint loading. Specifically, therefore the reductions in gluteus medius and gluteus minimus forces at the first hip joint force peak for the toe-in conditions and the reductions in rectus femoris forces at the second peak indicate that the alterations in joint loading were mediated via these changes in muscle kinetics. Hip joint loading is associated with arthritic degeneration (33), thus, the findings from this investigation indicate firstly that lateral orthoses do not appear to influence the risk from the parameters linked to hip joint OA and more importantly that whilst foot progression angles do influence hip joint loading at different parts of the stance phase they appear not to be able to do so in such a way that the risk from hip OA is attenuated. Although, it should be acknowledged that it is currently unknown which of the hip joint loading peaks are most strongly associated with OA initiation.

At the patellofemoral joint, the current investigation showed firstly that lateral orthoses did not influence patellofemoral joint loading, indicating that they may not be effective in mediating any statistical reductions in patellofemoral joint mechanics (2). However, this study did reveal using both SPM and discrete analyses that the neutral and lateral orthoses conditions were associated with statistical increases in both peak patellofemoral joint force/ stress and the impulse experienced throughout the stance phase. Taking into account the correlation analyses indicating that the vasti muscle group were most strongly associated with patellofemoral joint stress, and the associated increases in vastus intermedius forces noted in the neutral and lateral orthoses conditions, provides a mechanical explanation for the observed alterations in patellofemoral loading (21). Furthermore, given the proposed association between excessive joint stress and the aetiology of patellofemoral pain (2), the current investigation indicates that alterations in foot progression angles irrespective of direction (i.e. toe-in or toe-out conditions) are able to attenuate the magnitude of patellofemoral loading mechanisms linked to the aetiology of patellofemoral pain.

Examination of ankle joint kinetics showed via discrete analysis that the impulse of ankle joint loading across the stance phase was statistically larger in the toe-out and toe-in conditions in relation to the neutral and lateral orthoses walking modalities. The correlation analyses showed that the ankle joint force impulse was associated with lateral gastrocnemius, soleus and tibialis posterior muscle forces. Therefore, the associated increases in gastrocnemius and tibialis impulses in the toe-out conditions, explains the observed increases in ankle joint loading in this condition (21), although increases in the toe-in condition remain unresolved and may be due to patterns of muscle forces not considered in this study. Nonetheless, the association between ankle joint loading and the aetiology of ankle joint degeneration is well established (34), thus the current investigation shows that altering the habitual foot progression angle may increase the risk from ankle joint OA.

That this study utilized a musculoskeletal simulation-based procedure to quantify muscle and joint forces may serve as a limitation to the current investigation. Whist musculoskeletal simulation is considered an improvement over previous analyses of lateral orthoses and toe-in/toe-out gait using joint moments; it does depend on the underlying mathematical model and numerous mechanical assumptions are made in the construction of musculoskeletal simulation models. These predominately relate to the constrained rotational degrees of freedom at the lower extremity joints in addition to the lack of modelled soft tissues in particular around the knee joint itself, which may lead to incorrectly predicted muscle and joint forces (35).

In conclusion, although the mechanics of walking in lateral orthoses and with different foot progression angles have received previous research attention; there has yet to be a cumulative comparison of lower extremity joint loading, using a musculoskeletal simulation based approach. The present investigation adds to the current knowledge, by examining the effects of lateral orthoses as well as toe-in/ toe-out gait patterns on lower extremity joint loading. This investigation importantly showed that patellofemoral joint stress was greater in the neutral and lateral orthosis conditions and ankle joint loading larger in the toe-in and toe-out foot progression angle modalities. Furthermore, it was also shown that medial tibiofemoral impulse was statistically greater in the toe-in conditions. Therefore, the current investigation provides evidence that altering the foot progression angle to may attenuate the risk from patellofemoral disorders whilst simultaneously enhancing the risk from degenerative ankle pathologies. Similarly, adopting a toe-in foot progression angle may also increase the risk from medial tibiofemoral degeneration.

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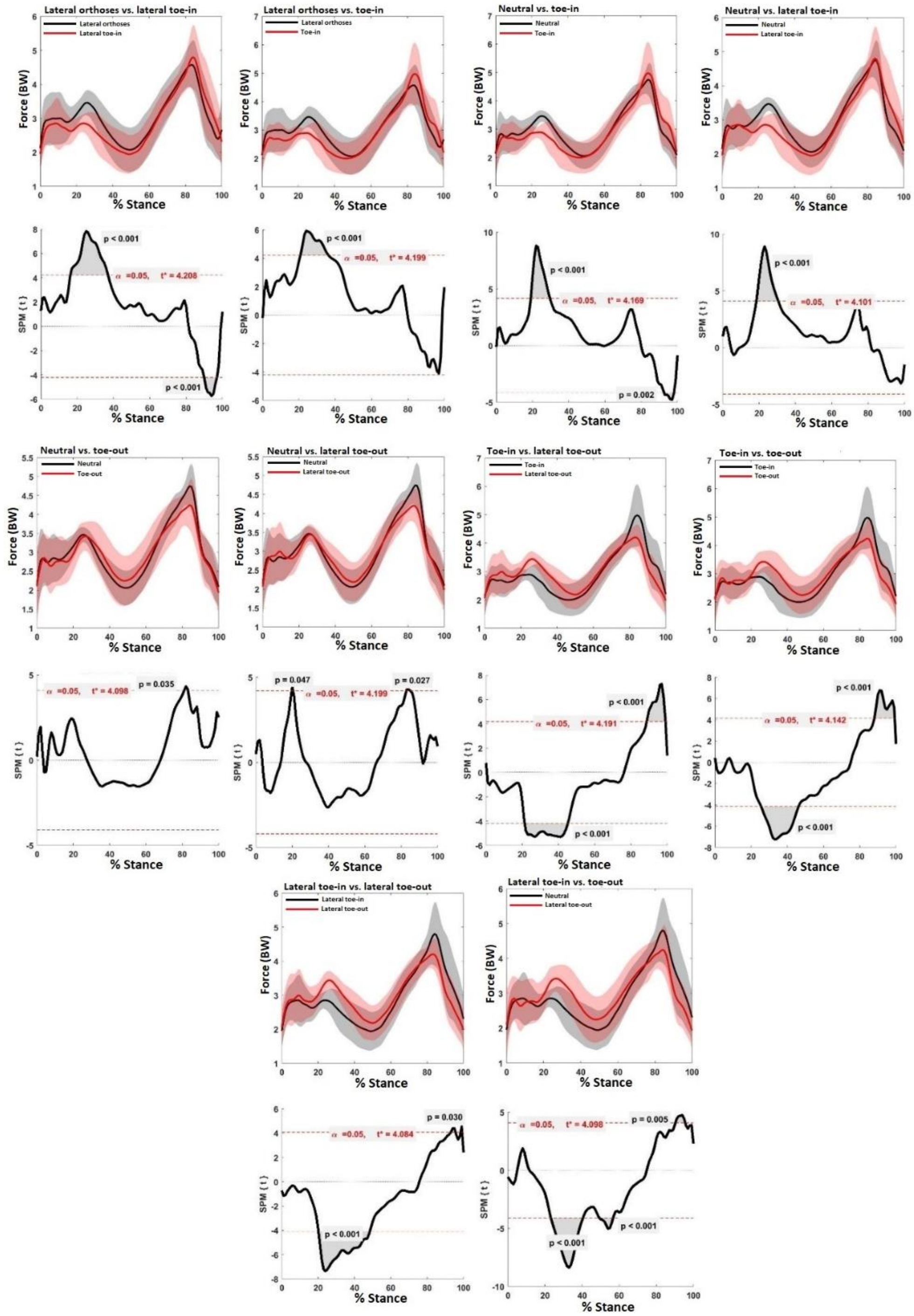

Figure 1. Comparison of hip forces across the stance phase, in different conditions. Positive SPM values indicate that values in the first above named condition exceed those in the other; SPM (t) denotes the t value and critical thresholds for statistical significance are denoted via the horizontal dotted lines.

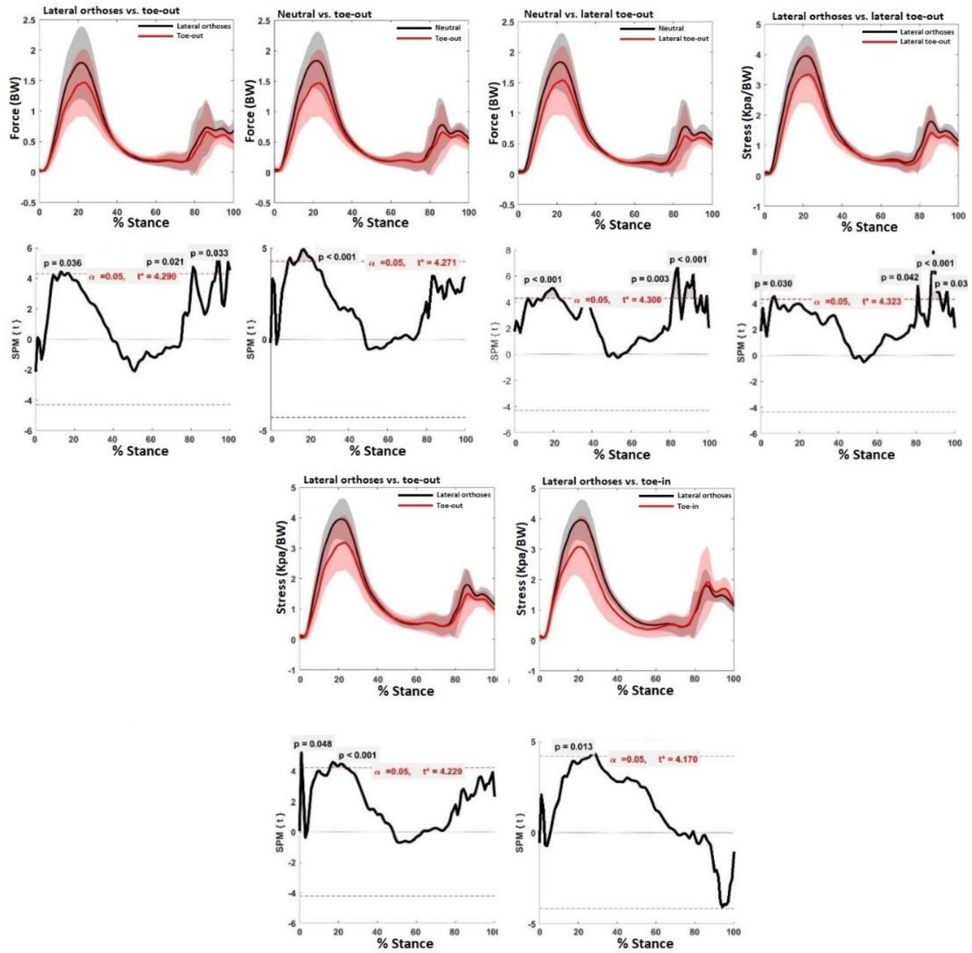

Figure 2. Comparison of patellofemoral forces/ stress across the stance phase, in different conditions. Positive SPM values indicate that values in the first above named condition exceed those in the other; SPM (t) denotes the t value and critical thresholds for statistical significance are denoted via the horizontal dotted lines.

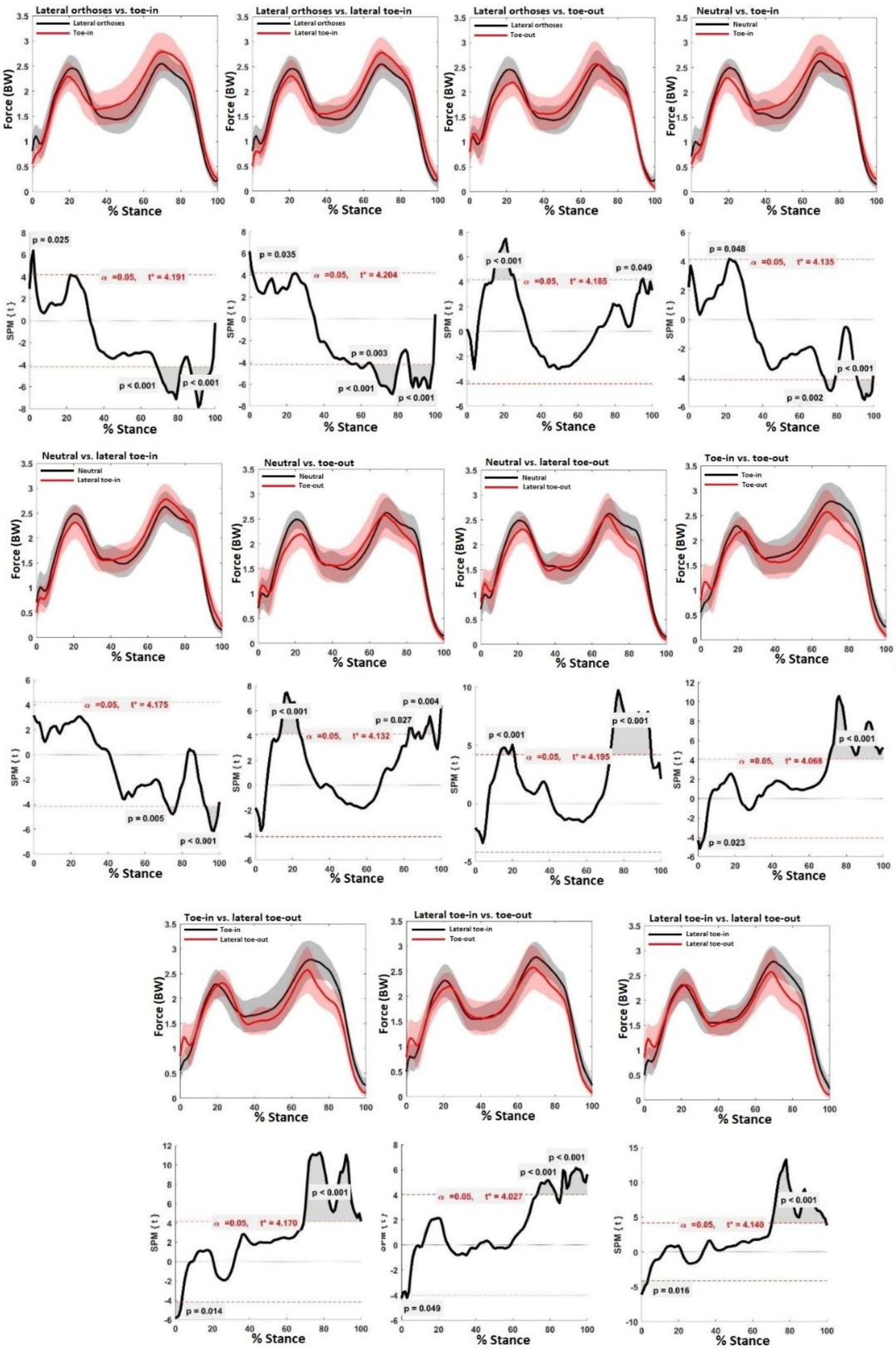

Figure 3. Comparison of medial tibiofemoral forces across the stance phase, in different conditions. Positive SPM values indicate that values in the first above named condition exceed those in the other; SPM (t) denotes the t value and critical thresholds for statistical significance are denoted via the horizontal dotted lines.

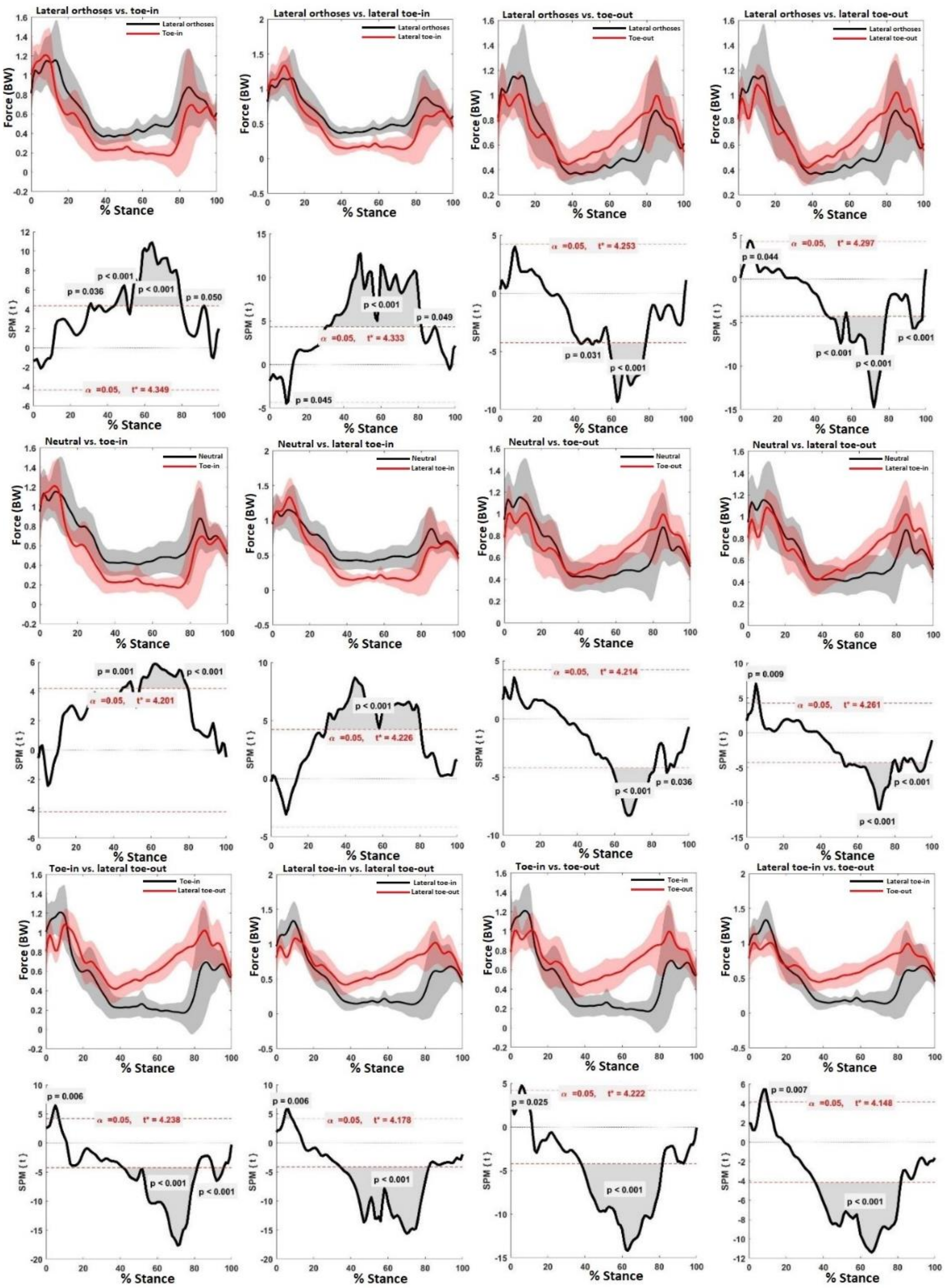

Figure 4. Comparison of lateral tibiofemoral forces across the stance phase, in different conditions. Positive SPM values indicate that values in the first above named condition exceed those in the other; SPM (t) denotes the t value and critical thresholds for statistical significance are denoted via the horizontal dotted lines.

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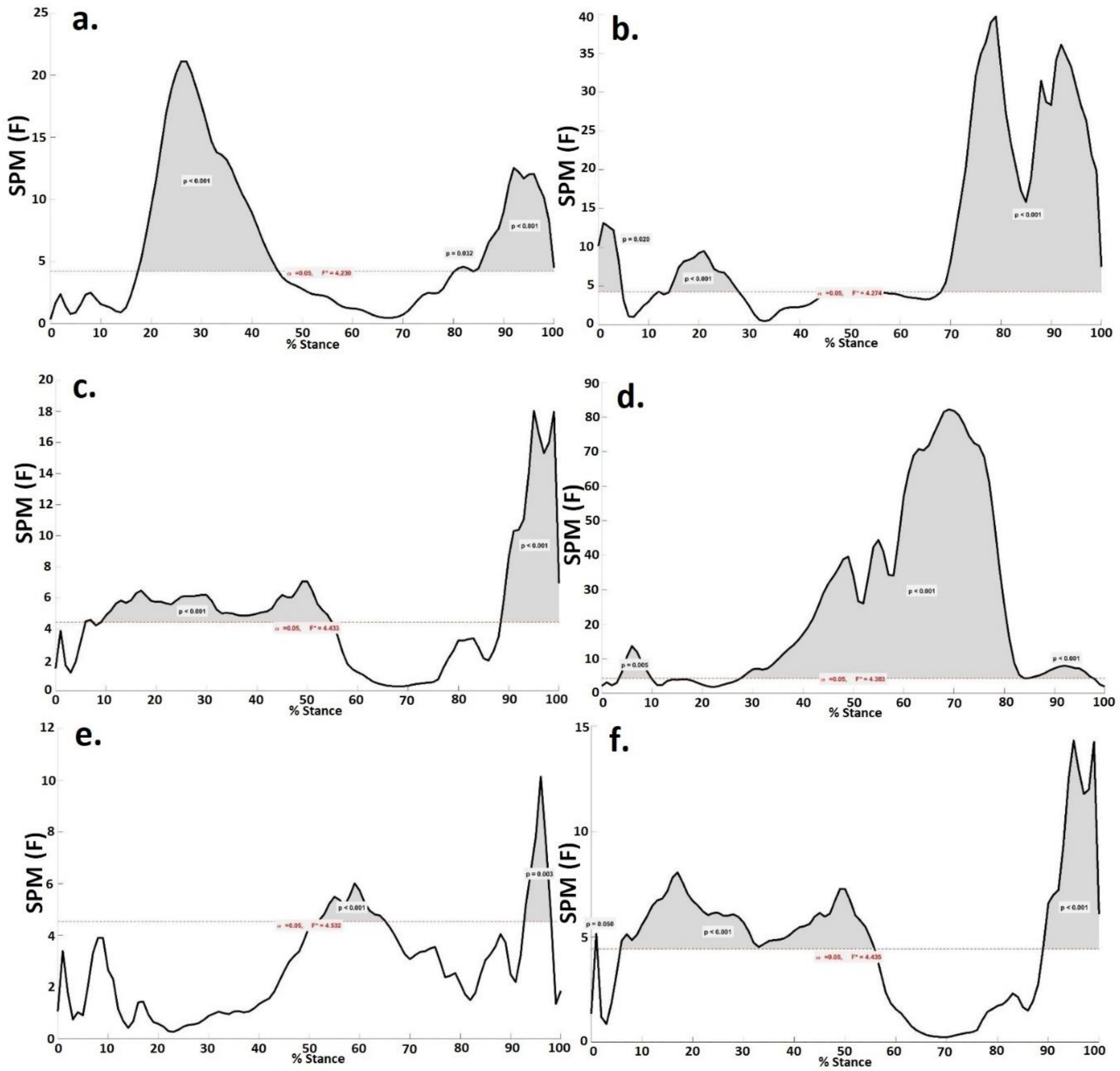

1. Comparison of the a. hip joint force, b. medial tibiofemoral force, c. patellofemoral force, d. lateral tibiofemoral force, e. ankle force and f. patellofemoral stress across the stance phase in all conditions. SPM (F) denotes the F value, and critical thresholds for statistical significance are denoted via the horizontal dotted line.

Table 1: Joint kinetics (Mean & standard deviations) as a function of orthoses and foot progression angle conditions.

	Lateral orthoses		Neutral		Toe-in		Lateral toe-in		Toe-out		Lateral toe-out	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD
Ankle force impulse (BW·s)	1.51	0.09	1.53	0.11	1.65	0.11	1.59	0.09	1.62	0.09	1.58	0.11
Lateral tibiofemoral impulse (BW·s)	0.40	0.07	0.40	0.08	0.31	0.03	0.30	0.08	0.45	0.05	0.45	0.07
First peak medial tibiofemoral force (BW)	2.50	0.26	2.55	0.19	2.38	0.26	2.35	0.32	2.28	0.30	2.35	0.28
Second peak medial tibiofemoral force (BW)	2.61	0.32	2.73	0.37	2.90	0.37	2.86	0.31	2.67	0.42	2.64	0.46
Medial tibiofemoral force impulse (BW·s)	1.07	0.10	1.07	0.09	1.20	0.17	1.15	0.09	1.09	0.16	1.05	0.11
Peak patellofemoral force (BW)	1.83	0.63	1.93	0.53	1.53	0.66	1.59	0.74	1.54	0.61	1.61	0.63
Peak patellofemoral stress (KPa/BW)	3.96	1.03	4.20	0.87	3.33	1.02	3.43	1.12	3.35	1.01	3.53	1.03
Patellofemoral stress impulse (KPa/BW·s)	0.89	0.14	0.90	0.14	0.80	0.23	0.79	0.22	0.80	0.15	0.78	0.13
First peak hip force (BW)	3.94	0.65	3.72	0.44	3.41	0.56	3.44	0.55	3.84	0.65	3.77	0.49
Second peak hip force (BW)	4.89	1.03	4.96	1.05	5.20	1.61	5.04	1.55	4.43	1.11	4.36	0.86

Table 2: Muscle forces and impulses (Mean & standard deviations) as a function of orthoses and foot progression angle conditions.

	Lateral orthoses		Neutral		Toe-in		Lateral toe-in		Toe-out		Lateral toe-out	
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD
Gluteus medius 2 first hip force peak (BW)	0.29	0.08	0.29	0.08	0.24	0.06	0.21	0.04	0.32	0.08	0.31	0.07
Gluteus medius 3 first hip force peak (BW)	0.30	0.14	0.30	0.08	0.27	0.13	0.21	0.10	0.33	0.16	0.33	0.13
Gluteus minimus 3 first hip force peak (BW)	0.10	0.03	0.10	0.02	0.09	0.02	0.08	0.02	0.11	0.03	0.09	0.02
Tensor fasciae latae first hip force peak (BW)	0.09	0.04	0.10	0.04	0.08	0.04	0.05	0.03	0.13	0.03	0.11	0.04
Rectus femoris first hip force peak (BW)	0.45	0.34	0.51	0.26	0.37	0.36	0.29	0.34	0.60	0.38	0.50	0.33
Gluteus medius 1 second hip force peak (BW)	0.47	0.17	0.46	0.11	0.35	0.14	0.38	0.15	0.41	0.11	0.42	0.13
Rectus femoris second hip force peak (BW)	1.03	0.63	1.05	0.46	1.24	0.85	1.10	0.79	0.81	0.73	0.78	0.67
Vastus intermedius first medial tibiofemoral force peak (BW)	0.48	0.20	0.50	0.10	0.42	0.16	0.44	0.18	0.38	0.18	0.42	0.14
Vastus lateralis first medial tibiofemoral force peak (BW)	0.86	0.36	0.92	0.19	0.74	0.31	0.78	0.32	0.68	0.33	0.75	0.26
Vastus medialis first medial tibiofemoral force peak (BW)	0.41	0.16	0.42	0.09	0.37	0.14	0.37	0.15	0.31	0.15	0.35	0.12
Rectus femoris first medial tibiofemoral force peak (BW)	0.33	0.35	0.33	0.21	0.24	0.24	0.22	0.23	0.43	0.34	0.33	0.24
Lateral gastrocnemius second medial tibiofemoral force peak (BW)	0.28	0.06	0.29	0.04	0.26	0.08	0.26	0.07	0.33	0.07	0.32	0.06
Vastus intermedius patellofemoral force peak (BW)	0.49	0.17	0.51	0.09	0.43	0.16	0.46	0.17	0.40	0.15	0.43	0.14
Lateral gastrocnemius impulse (BW·s)	0.06	0.01	0.07	0.01	0.07	0.02	0.06	0.01	0.08	0.02	0.07	0.01
Medial gastrocnemius impulse (BW·s)	0.28	0.04	0.28	0.04	0.32	0.09	0.30	0.04	0.33	0.06	0.31	0.05
Tibialis anterior impulse (BW·s)	0.04	0.02	0.04	0.01	0.04	0.02	0.04	0.01	0.04	0.01	0.05	0.02
Tibialis posterior impulse (BW·s)	0.04	0.02	0.07	0.05	0.03	0.03	0.03	0.03	0.15	0.08	0.15	0.08

Table 3: Correlations between joint kinetic parameters and foot progression angles with associated Bayes factors.

	r	**BF**
Lateral tibiofemoral impulse	0.73	1.78E+11
Second peak medial tibiofemoral force	-0.37	8.21
Medial tibiofemoral force impulse	-0.34	10.12

Table 4: Correlations between joint kinetic parameters and muscles forces/impulses with associated Bayes factors.

First hip force peak			**Second hip force peak**		
	r	**BF**		**r**	**BF**
Adductor magnus 1	0.56	156069.21	Adductor magnus 1	0.61	4451247.16
Adductor magnus 2	0.56	112579.04	Adductor magnus 2	0.55	50498.31
Adductor magnus 3	0.50	4737.95	Adductor magnus 3	0.40	72.08
Gluteus maximus 1	0.59	863819.92	Rectus femoris	0.96	2.75E+40
Gluteus maximus 2	0.58	471381.57	Gluteus medius 2	-0.53	16994.49
Gluteus maximus 3	0.52	14173.44	Gluteus medius 3	-0.49	2226.24
Gluteus medius 2	0.33	7.32	Gluteus minimus 1	0.82	3.29E+17
Gluteus medius 3	0.49	2590.80	Gluteus minimus 2	0.82	1.54E+17
Gluteus medius 2	0.38	32.49	Gluteus minimus 3	0.59	799133.16
Gluteus medius 3	0.56	144058.66	Tensor fasciae latae	0.76	1.08E+13
First medial tibiofemoral peak			**Second medial tibiofemoral peak**		
	r	**BF**		**r**	**BF**
Vastus intermedius	0.70	5.85E+09	Lateral gastrocnemius	0.30	3.23
Vastus lateralis	0.70	5.55E+09	Medial gastrocnemius	0.72	8.45E+10
Vastus medialis	0.68	1.09E+09			
Peak patellofemoral stress			**Peak lateral tibiofemoral force**		
	r	**BF**		**r**	**BF**
Vastus intermedius	0.71	1.64E+10	Vastus lateralis	0.31	4.30
Vastus lateralis	0.71	2.29E+10	Biceps femoris long head	0.30	3.29
Vastus medialis	0.70	9.66E+09			
Ankle impulse					
	r	**BF**			
Lateral gastrocnemius	0.54	37890.15			
Medial gastrocnemius	0.49	2787.68			
Soleus	0.45	486.95			
Tibialis posterior	0.31	3.59			