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

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11 **Keywords:** biomechanics, orthoses, toe-out, toe-in, gait.

12 **Abstract**

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

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

21 **RESULTS:** This investigation showed that peak patellofemoral joint stress was greater in the
22 neutral (3.96 KPa/BW) and lateral orthoses (4.20 KPa/BW) conditions compared to toe-in (3.33
23 KPa/BW), lateral toe-in (3.43 KPa/BW), toe-out (3.35 KPa/BW) and lateral toe-out (3.53
24 KPa/BW) and ankle joint impulse larger in the toe-in (1.65BW·s) and toe-out (1.62BW·s) foot

25 progression angle modalities compared to neutral (1.51BW·s) and lateral orthoses (1.53BW·s).
26 Furthermore, it was also shown that medial tibiofemoral impulse was statistically greater in the
27 toe-in (1.20BW·s) and lateral toe-in (1.15BW·s) conditions compared to neutral (1.07BW·s),
28 lateral orthoses (1.07BW·s), toe-out (1.09BW·s) and lateral toe-out (1.05BW·s).

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

33

34 **Introduction**

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

42

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

50

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

61

62 A further conservative modality for the reduction of the knee adduction moment is alterations
63 to individuals walking gait pattern (1). Changing the foot progression angle has two variants;
64 toe-out and toe-in gait, which unlike other gait modification techniques appear to be sustainable
65 up to 10 weeks (17). Toe-in (represented by internal rotation of the foot) and toe-out
66 (characterized by external rotation of the foot) gait patterns are similarly designed to influence
67 the knee adduction moment by moving the centre of pressure mediolaterally (17). Typically
68 toe-in gait patterns have been shown to reduce the magnitude of the first peak in the knee
69 adduction moment time curve (1; 18-19), yet toe-out gait has correspondingly been found to
70 reduce the second knee adduction moment peak (1; 18-20). However, conversely Jenkyn et al.,
71 (2008) showed that that toe-out walking reduced both the first and second peak of the knee
72 adduction moment curve. Similarly, both Van den Noort et al., (18) and Khan et al., (1) showed
73 that the knee adduction moment impulse was reduced with a toe-in gait pattern, yet other

74 investigations have shown that there is no effect of altering the foot progression angle on this
75 parameter (17).

76

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

89

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

97

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

105

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

109

110 **Methods**

111 *Participants*

112 Fifteen male participants (age 31.73 ± 4.96 years, height 1.72 ± 0.06 m and body mass
113 69.31 ± 9.92 kg and BMI = 23.45 ± 2.78 kg/m²). volunteered to take part in this study. All
114 participants were free from pathology at the time of data collection and provided written
115 informed consent, in accordance with the principles outlined in the Declaration of Helsinki.
116 The inclusion criteria for the subjects included healthy adults, aged 20–45 years and having a
117 BMI of less than 30 kg/m² (1). The participants were excluded on the basis of any
118 musculoskeletal disorder, previous knee surgery or inability to adopt the novel gait pattern. The
119 procedure utilized for this investigation was approved; by a university ethical committee
120 (STEMH 1013).

121

122 *Experimental orthoses*

123 Commercially available full-length orthoses (Slimflex Simple, High Density, Full Length,
124 Algeos UK) were examined in the current investigation. The orthoses were made from
125 ethylene-vinyl acetate with a shore A rating of 65 and had a heel thickness of 11 mm including
126 the additional wedge. The orthoses were featured a 5° lateral wedge configuration which
127 spanned the full length of the device. To ensure consistency each participant wore the same
128 footwear (Asics, Gel Patriot 6). The experimental footwear had a mean mass of 0.265 kg, heel
129 thickness of 22 mm and heel drop of 10 mm.

130

131 *Foot progression angles*

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

144

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

151

152 *Procedures*

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

166

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

176

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

182

183 *Processing*

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

189

190 Data during the stance phase were exported from Visual 3D into OpenSim 3.3 software
191 (Simtk.org). A validated musculoskeletal model with 12 segments, 19 degrees of freedom and
192 92 musculotendon actuators (27) was used to estimate lower extremity joint forces. The model
193 featured the same segments and muscle tendon units as the standard Gait2392 Opensim model
194 ([https://simtk-](https://simtk-confluence.stanford.edu:8443/display/OpenSim/Gait+2392+and+2354+Models?preview=/376103/3736767/Gait%202392%20vs.%20Gait%202354.pdf)
195 [confluence.stanford.edu:8443/display/OpenSim/Gait+2392+and+2354+Models?preview=/3](https://simtk-confluence.stanford.edu:8443/display/OpenSim/Gait+2392+and+2354+Models?preview=/376103/3736767/Gait%202392%20vs.%20Gait%202354.pdf)
196 [376103/3736767/Gait%202392%20vs.%20Gait%202354.pdf](https://simtk-confluence.stanford.edu:8443/display/OpenSim/Gait+2392+and+2354+Models?preview=/376103/3736767/Gait%202392%20vs.%20Gait%202354.pdf)) but the tibiofemoral joint was
197 separated into medial and lateral compartments to allow joint reaction analyses at these areas
198 separately and the model also featured a patella. The model was firstly scaled for each
199 participant to account for the anthropometrics of each individual. Then as muscle forces are
200 the main determinant of joint compressive forces (21), muscle kinetics were quantified using
201 static optimization. **The static optimization simulation process calculates the muscle forces**
202 **required in order to recreate the experimentally measured motions.** Compressive ankle, medial/
203 lateral tibiofemoral, patellofemoral and hip joint forces were calculated via the joint reaction
204 analyses function within OpenSim using the muscle forces generated from the static
205 optimization process as inputs. The joint reaction analysis function in OpenSim calculates the
206 joint loads transferred between two contacting bodies, about the joint centre location identified
207 during the static trial (28). In the current investigation, joint forces were normalized by dividing
208 by each participants body weight (BW). Compressive hip joint forces were representative of
209 the contact forces between the femur and acetabular cartilage, tibiofemoral forces between the
210 medial/ lateral tibial and femoral cartilage, patellofemoral joint forces between the femur and
211 patella cartilage and ankle joint forces between the tibia and talar cartilage. Patellofemoral joint
212 stress (KPa/BW) was also quantified by dividing the patellofemoral joint reaction force by the
213 patellofemoral contact area. Patellofemoral contact areas were obtained by fitting a polynomial
214 curve to the sex specific data of Besier et al., (29). From the above processing, peak normalized

215 ankle force, peak lateral tibiofemoral force, and peak patellofemoral force/ stress during the
216 stance phase were extracted for statistical analyses. In addition, as the hip and medial
217 tibiofemoral joint force curves exhibit a double peaked pattern, the normalized values at the
218 first and second peak for these parameters were extracted for analysis.

219

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

225

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

231

232 *Statistical analyses*

233 Following data processing, compressive joint forces (hip, patellofemoral, ankle, medial/ lateral
234 tibiofemoral), during the stance phase were exported and temporally normalized using linear
235 interpolation to 101 data points for statistical analysis using Statistical parametric mapping
236 (SPM). In agreement with Pataky et al. (30), SPM was implemented in a hierarchical manner,
237 analogous to one-way repeated measures ANOVA with post-hoc t-tests. Therefore, the entire
238 data-set was examined first, and if a statistical main effect was reached then post-hoc tests

239 comparing individual conditions were conducted on each component separately. For discrete
240 parameters (peak joint forces, joint impulse, joint instantaneous load rates, muscle forces and
241 muscle force impulses), means and standard deviations were calculated for each condition.
242 Differences in discrete biomechanical parameters were examined using Bayesian one-way
243 repeated measures ANOVA with default prior scales using JASP software 0.10.2 (31). In the
244 event of a main effect, post-hoc Bayesian paired t-tests were conducted between each
245 condition. Similarly, relationships between 1. discrete joint loads and foot progression angle
246 and 2. discrete joint loads and muscle forces at the instances of peak joint loads/ muscle force
247 impulses across all conditions were examined using Bayesian Pearson's correlation analyses
248 using SPSS 27.0 software (SPSS, IBM). Bayesian factors (BF) were used to explore the extent
249 to which the data supported the alternative (H_1) hypothesis. Bayes factors throughout were
250 interpreted in accordance with the recommendations of Jeffreys, (32), with values ≥ 3 indicating
251 sufficient evidence in support of H_1 . In the interests of conciseness only variables that present
252 with Bayes factors ≥ 3 and SPM analyses showing statistical significance will be presented.

253

254 **Results**

255 *Foot progression angles*

256 The mean \pm SD of foot progression angles were: neutral = $0.68 \pm 2.70^\circ$, lateral orthoses = 1.47
257 $\pm 2.85^\circ$, toe-out = $12.38 \pm 4.42^\circ$, lateral toe-out = $12.44 \pm 3.58^\circ$, toe-in = $-10.63 \pm 2.82^\circ$ and
258 lateral toe-in = $-10.54 \pm 2.37^\circ$.

259

260 *Discrete analyses – joint loading*

261

@@@Table 1 near here@@@

262 For the first peak of the hip force there was decisive evidence of a main effect (BF =
263 129890.84). Post-hoc analyses showed that values were larger in the lateral orthoses (BF =
264 42.84), neutral (BF = 35.68), toe-out (BF = 22.47), lateral toe-out (BF = 10.91) conditions
265 compared to toe-in. Furthermore, values were also larger in the lateral orthoses (BF = 149.13),
266 neutral (BF = 7.06), toe-out (BF = 3.67), lateral toe-out (BF = 22.12) conditions compared to
267 lateral toe-in (Table 1). For the second peak of the hip force there was decisive evidence of a
268 main effect (BF = 24.93). Post-hoc analyses showed that values were larger in the lateral
269 orthoses (BF = 9.28), neutral (BF = 7.37), toe-in (BF = 5.93) and lateral toe-in (BF = 3.04)
270 conditions compared to toe-out. Furthermore, values were also larger in the lateral orthoses
271 (BF = 16.21), neutral (BF = 32.86) and lateral toe-out (BF = 3.16) conditions compared to
272 lateral toe-out (Table 1).

273

274 For the first peak of the medial tibiofemoral force there was decisive evidence of a main effect
275 (BF = 298.86). Post-hoc analyses showed that values were larger in the neutral condition
276 compared to toe-in (BF = 5.82), lateral toe-in (BF = 4.73), toe-out (BF = 229.47) and lateral
277 toe-out (BF = 20.24). In addition, values for this parameter were larger in lateral orthoses
278 compared to lateral toe-in (BF = 4.08), toe-out (BF = 14.12) and lateral toe-out (BF = 3.54)
279 (Table 1). For the second peak of the medial tibiofemoral force there was also decisive evidence
280 of a main effect (BF = 8543.01). Post-hoc analyses showed that values were larger in the toe-
281 in condition compared to lateral orthoses (BF = 248.62), neutral (BF = 4.30), toe-out (BF =
282 5.27) and lateral toe-out (BF = 15.67). In addition, values for this parameter were larger in the
283 lateral toe-in condition compared to lateral orthoses (BF = 77.03) and lateral toe-out (BF =
284 3.27) (Table 1). For the medial tibiofemoral force impulse there was decisive evidence of a
285 main effect (BF = 499513.34). Post-hoc analyses showed that values were larger in the toe-in
286 condition compared to lateral orthoses (BF = 14.72), neutral (BF = 13.71), toe-out (BF =

287 175.33) and lateral toe-out (BF = 783.24). In addition, values for this parameter were larger in
288 the lateral toe-in condition compared to lateral orthoses (BF = 621.47) and lateral toe-out (BF
289 = 236.42) (Table 1).

290

291 For the lateral tibiofemoral force impulse there was decisive evidence of a main effect (BF =
292 1442262604595.22). Post-hoc analyses showed that values were larger in the lateral orthoses
293 condition compared to toe-in (BF = 225.77) and lateral toe-in (BF = 66.20) and in the neutral
294 compared to toe-in (BF = 31.14) and lateral toe-in (BF = 25.76). In addition, values for this
295 parameter were larger in the toe-out condition compared to lateral orthoses (BF = 4.63), neutral
296 (BF = 7.74), toe-in (BF = 10906.26) and lateral toe-in (133.70) and also in the lateral toe-out
297 compared to lateral orthoses (BF = 11.82), neutral (BF = 34.14), toe-in (BF = 2393.76) and
298 lateral toe-in (1151.05) (Table 1).

299

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

305

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

310 19.19) (Table 1). In addition, for the patellofemoral stress impulse there was substantial
311 evidence of a main effect (BF = 3.87). Post-hoc analyses showed that values were larger in the
312 neutral condition compared to toe-out (BF = 31.48) and lateral toe-out (BF = 212.27). In
313 addition, values for this parameter were larger in lateral orthoses compared to toe-out (BF =
314 4.88) and lateral toe-out (BF = 29.24) (Table 1).

315

316 For the ankle force impulse there was decisive evidence of a main effect (BF = 12448.22). Post-
317 hoc analyses showed that values were larger in the toe-out condition compared to neutral (BF
318 = 8.43) and lateral orthoses (BF = 152.30). In addition, values for this parameter were larger in
319 the toe-in condition compared to neutral (BF = 52.45) and lateral orthoses (BF = 184.65) and
320 also in the lateral toe-in condition compared to lateral orthoses (BF = 34.73) (Table 1).

321

322 Discrete analyses – muscles forces

323

@@@Table 2 near here@@@

324 For the gluteus medius 2 at the instance of the first peak of the hip force there was decisive
325 evidence of a main effect (BF = 381404.40). Post-hoc analyses showed that values were larger
326 in the lateral orthoses (BF = 10.46), neutral (BF = 755.02), toe-out (BF = 10.78), lateral toe-
327 out (BF = 57.35) conditions compared to toe-in. Furthermore, values were also larger in the
328 lateral orthoses (BF = 87.33), neutral (BF = 19.43), toe-out (BF = 98.26), lateral toe-out (BF =
329 251.58) conditions compared to lateral toe-in (Table 2). Similarly, for the gluteus medius 3 at
330 the instance of the first peak of the hip force there was decisive evidence of a main effect (BF
331 = 732.05). Post-hoc analyses showed that values were larger in the lateral orthoses (BF =
332 103.36), neutral (BF = 10.60), toe-out (BF = 28.17), lateral toe-out (BF = 124.47) conditions

333 compared to lateral toe-in (Table 2). For the gluteus minimus 3 at the instance of the first peak
334 of the hip force there was substantial evidence of a main effect (BF = 6.02). Post-hoc analyses
335 showed that values were larger in the lateral orthoses (BF = 46.62), neutral (BF = 10.02), toe-
336 out (BF = 4.89), lateral toe-out (BF = 6.48) conditions compared to lateral toe-in (Table 2). For
337 the tensor fasciae latae at the instance of the first peak of the hip force there was decisive
338 evidence of a main effect (BF = 14450.22). Post-hoc analyses showed that values were larger
339 in the lateral orthoses (BF = 30.00), neutral (BF = 22.22), toe-out (BF = 4102.04), lateral toe-
340 out (BF = 47.70) conditions compared to lateral toe-in. Furthermore, values were also larger in
341 the toe-out compared to lateral orthoses (BF = 14.80) and toe-in (BF = 48.83) conditions (Table
342 2). For the rectus femoris at the instance of the first peak of the hip force there was decisive
343 evidence of a main effect (BF = 721.75). Post-hoc analyses showed that values were larger in
344 the lateral orthoses (BF = 14.97), toe-out (BF = 777619.80) and lateral toe-out (BF = 12.80)
345 conditions compared to lateral toe-in. Furthermore, values were also larger in the toe-out
346 compared to lateral orthoses (BF = 16.12) and toe-in (BF = 518.30) conditions (Table 2).

347

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

352

353 For the vastus intermedius at the instance of the first peak of the medial tibiofemoral force,
354 there was substantial evidence of a main effect (BF = 5.72). Post-hoc analyses showed that
355 values were larger in the lateral orthoses (BF = 82.13) and neutral (BF = 5.76) conditions
356 compared to toe-out (Table 2). For the vastus lateralis at the instance of the first peak of the

357 medial tibiofemoral force, there was substantial evidence of a main effect (BF = 8.22). Post-
358 hoc analyses showed that values were greater in the lateral orthoses (BF = 25.35) and neutral
359 (BF = 6.58) conditions compared to toe-out and also in the neutral compared to toe-in (BF =
360 3.16) and lateral toe-out (BF = 4.92) (Table 2). For the vastus medialis at the instance of the
361 first peak of the medial tibiofemoral force, there was substantial evidence of a main effect (BF
362 = 4.29). Post-hoc analyses showed that values were larger in the lateral orthoses (BF = 186.62)
363 and neutral (BF = 5.64) conditions compared to toe-out (Table 2). For the rectus femoris at the
364 instance of the first peak of the medial tibiofemoral force, there was strong evidence of a main
365 effect (BF = 14.77). Post-hoc analyses showed that values were larger in the lateral orthoses
366 (BF = 58.18), toe-in (BF = 109.18) and lateral toe-in (BF = 79.75) conditions compared to toe-
367 out and also in the toe-in (BF = 7.40) compared to lateral toe-out condition (Table 2).

368

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

375

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

380

381 For the lateral gastrocnemius impulse there was decisive evidence of a main effect (BF =
382 36188.95). Post-hoc analyses showed that values were greater in the toe-out compared to lateral
383 orthoses (BF = 276.36), neutral (BF = 24.91) and toe-in (BF = 38.24) also in the lateral toe-out
384 compared to lateral orthoses (BF = 80189.72), neutral (BF = 33.57) and toe-in (BF = 154.09)
385 conditions. There was also very strong evidence of a main effect for medial gastrocnemius
386 impulse (BF = 37.47). Post-hoc analyses showed that values were greater in the toe-out
387 compared to lateral orthoses (BF = 18.87) and neutral (BF 5.22) and also in the lateral toe-out
388 compared to lateral (BF = 99.06) and neutral (BF = 35.27) conditions. There was very strong
389 evidence of a main effect for tibialis anterior impulse (BF = 49.40). Post-hoc analyses showed
390 that values were larger in the toe-out condition compared to toe-in (BF = 17.27) and lateral toe-
391 in (BF 502.53). Finally, there was also decisive evidence of a main effect for tibialis posterior
392 impulse (BF = 645123.12). Post-hoc analyses showed that values were greater in the toe-out
393 condition compared to lateral orthoses (BF = 66.82), neutral (BF = 8.69) toe-in (BF = 656.32)
394 and lateral toe-in (BF = 625.85) and also in the lateral orthoses (BF = 64.81), neutral (BF =
395 12.78) toe-in (BF = 580.43) and lateral toe-in (BF = 829.33) conditions.

396

397 Statistical parametric mapping

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

402

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

412

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

420

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

428

429 For lateral tibiofemoral force, lateral orthoses were associated with increased lateral loading
430 during early-late stance compared to toe-in and lateral toe-in (Figure 4ab). In addition, lateral
431 orthoses were associated with increased loading during early stance compared to lateral toe-
432 out yet reduced loading during mid-late stance compared to toe-out and lateral toe-out (Figure
433 4cd). Similarly, the neutral condition was associated with increased lateral loading during
434 early-late stance compared to toe-in and lateral toe-in (Figure 4). In addition, the neutral
435 condition was associated with increased loading during early stance compared to lateral toe-
436 out yet reduced loading during late stance compared to toe-out and lateral toe-out (Figure 4).
437 Furthermore, the toe-in condition and lateral toe-in conditions were associated with increased
438 loading in early stance but enhanced loading during late stance compared to toe-out and lateral
439 toe-out (Figure 4).

440

441 Correlation analyses

442

@@@Table 3 near here@@@

443

@@@Table 4 near here@@@

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

448

449 **Discussion**

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

456

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

475

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

488

489 Much like at the medial tibiofemoral compartment, both the discrete and SPM based analyses
490 at the hip joint showed a contradictory pattern of statistical differences. Firstly, in the neutral
491 and lateral orthoses conditions as well as the toe-out and lateral toe-out conditions, the smaller
492 first peak of the hip joint contact force was enhanced compared to the toe-in and lateral toe-in
493 conditions. Conversely, it was also revealed that neutral and lateral orthoses conditions as well
494 as the toe-in and lateral toe-in modalities were associated with increases in the larger second
495 peak of the hip joint contact force in relation to the toe-out conditions. Importantly muscle
496 forces are the primary determinant of joint loading (21). Therefore the observations from the
497 correlational analyses showing that the first peak was best associated with adductor magnus
498 and gluteal muscle forces and the second peak with adductor magnus, gluteal, rectus femoris

499 and tensor fasciae latae provide insight into the mechanisms responsible for the aforementioned
500 changes in hip joint loading. Specifically, therefore the reductions in gluteus medius and
501 gluteus minimus forces at the first hip joint force peak for the toe-in conditions and the
502 reductions in rectus femoris forces at the second peak indicate that the alterations in joint
503 loading were mediated via these changes in muscle kinetics. Hip joint loading is associated
504 with arthritic degeneration (33), thus, the findings from this investigation indicate firstly that
505 lateral orthoses do not appear to influence the risk from the parameters linked to hip joint OA
506 and more importantly that whilst foot progression angles do influence hip joint loading at
507 different parts of the stance phase they appear not to be able to do so in such a way that the risk
508 from hip OA is attenuated. Although, it should be acknowledged that it is currently unknown
509 which of the hip joint loading peaks are most strongly associated with OA initiation.

510

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

523 are able to attenuate the magnitude of patellofemoral loading mechanisms linked to the
524 aetiology of patellofemoral pain.

525

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

537

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

546

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

560

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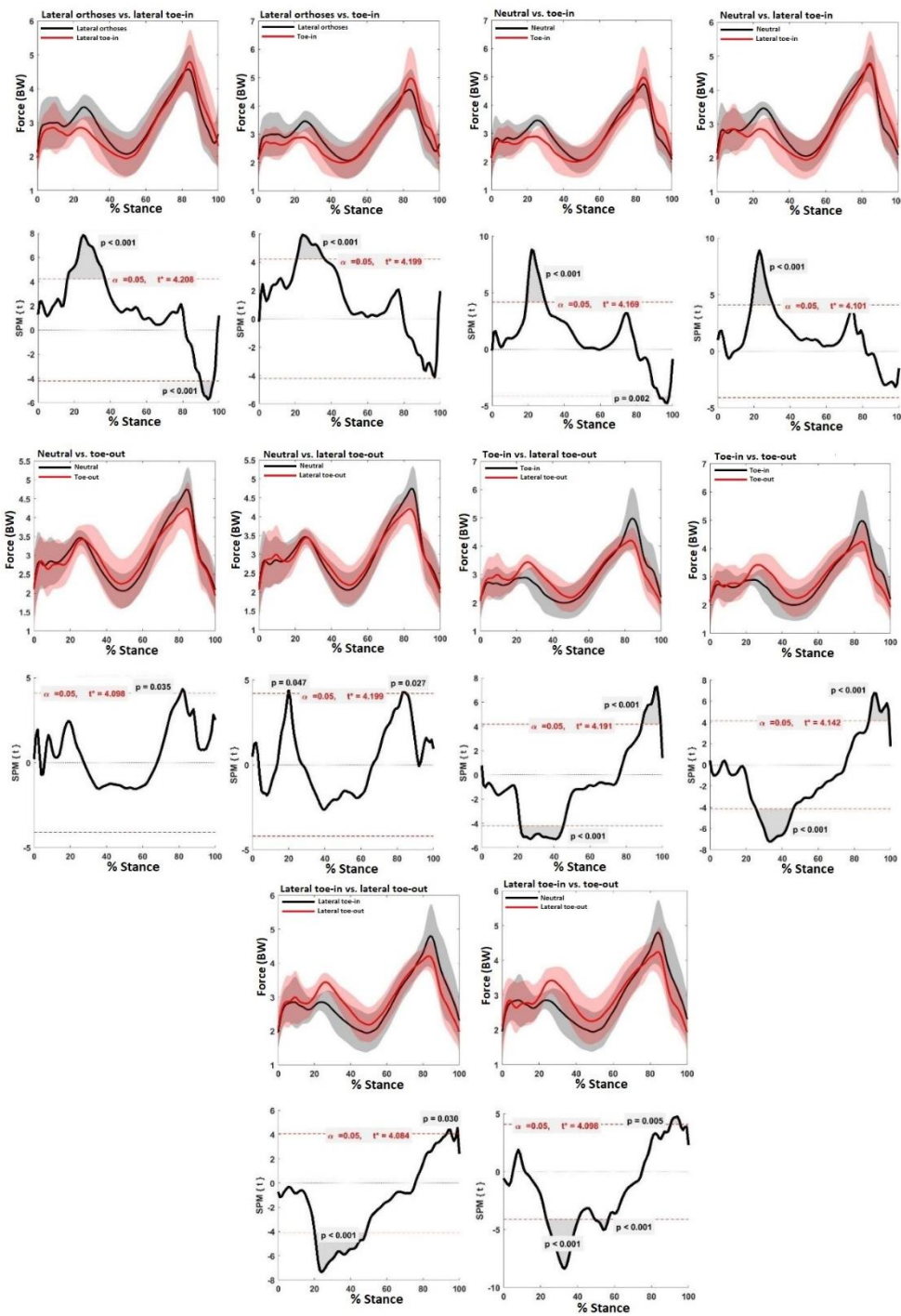
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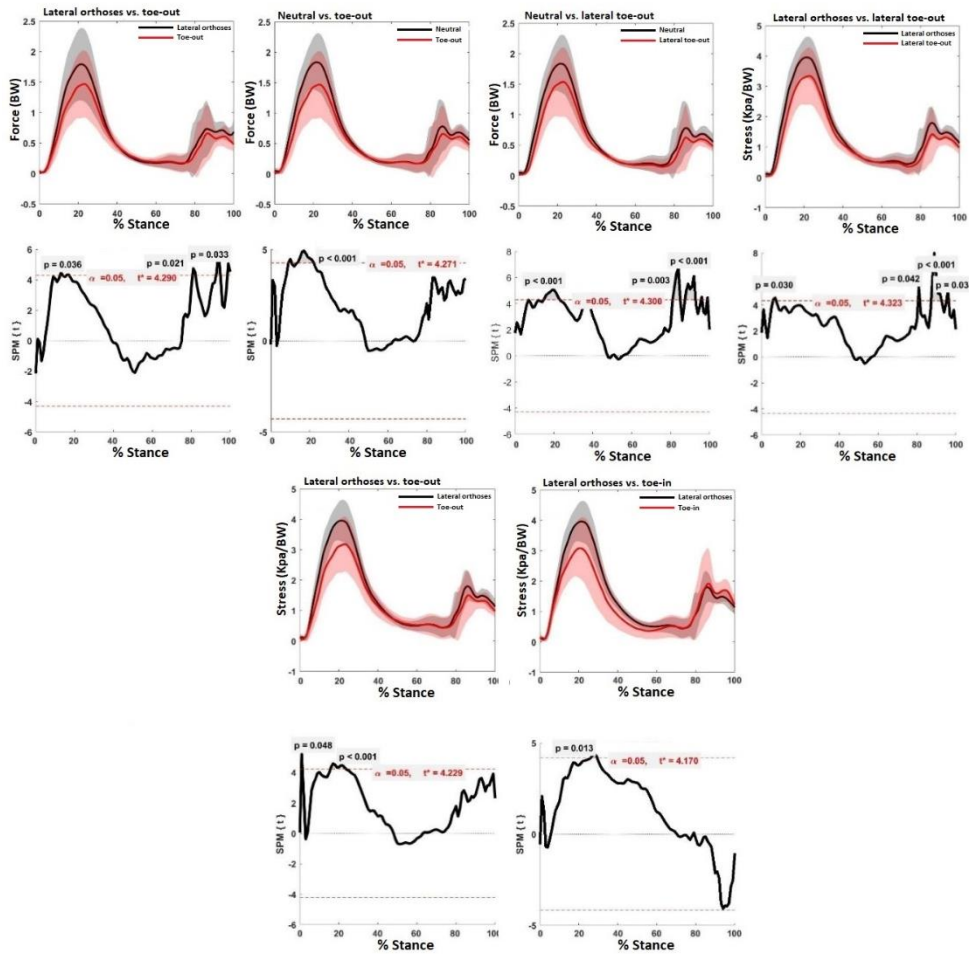
665 **List of figures**



666

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

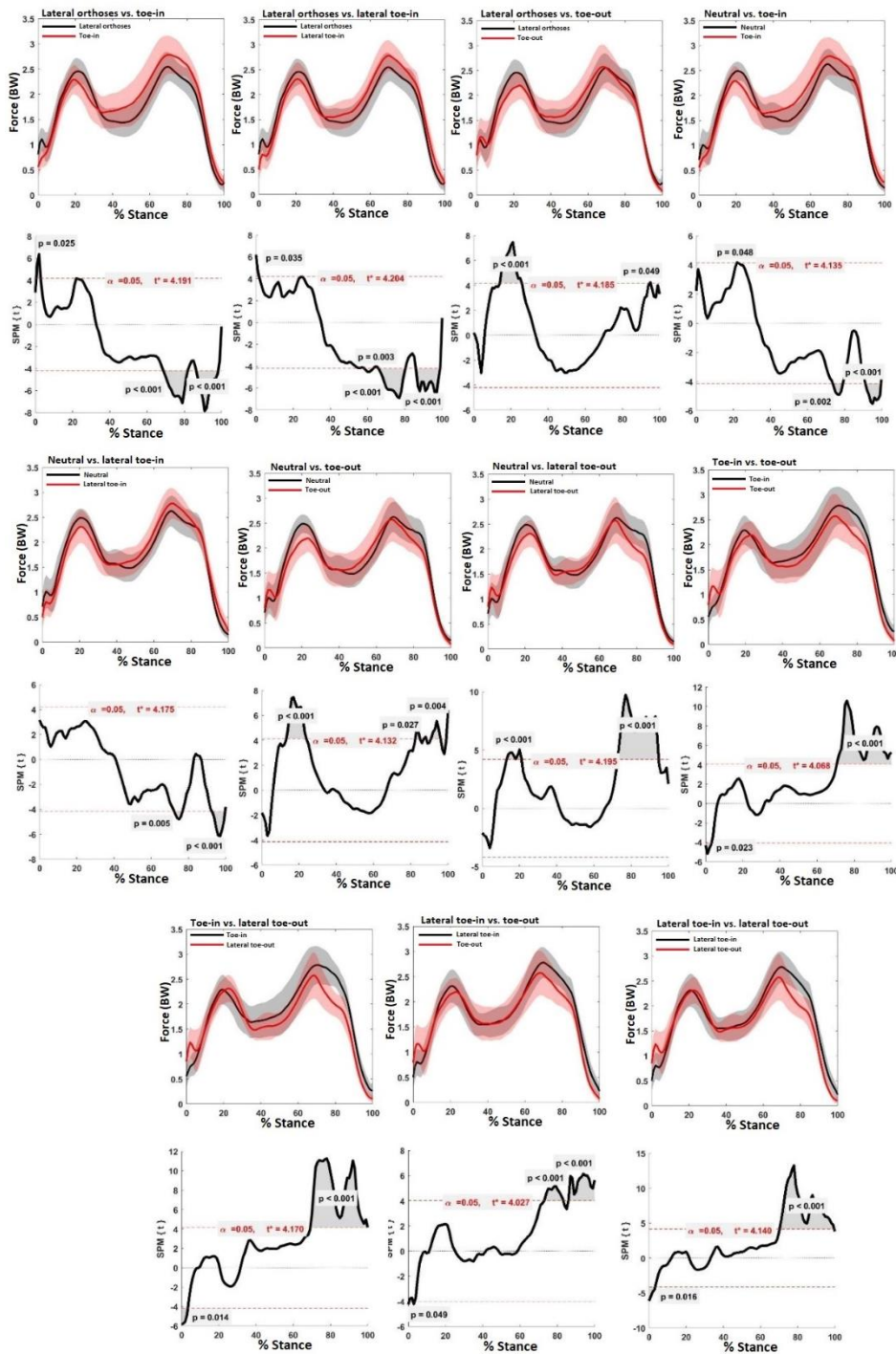
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673 Figure 2. Comparison of patellofemoral forces/ stress across the stance phase, in different
 674 conditions. Positive SPM values indicate that values in the first above named condition exceed
 675 those in the other; SPM (t) denotes the t value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

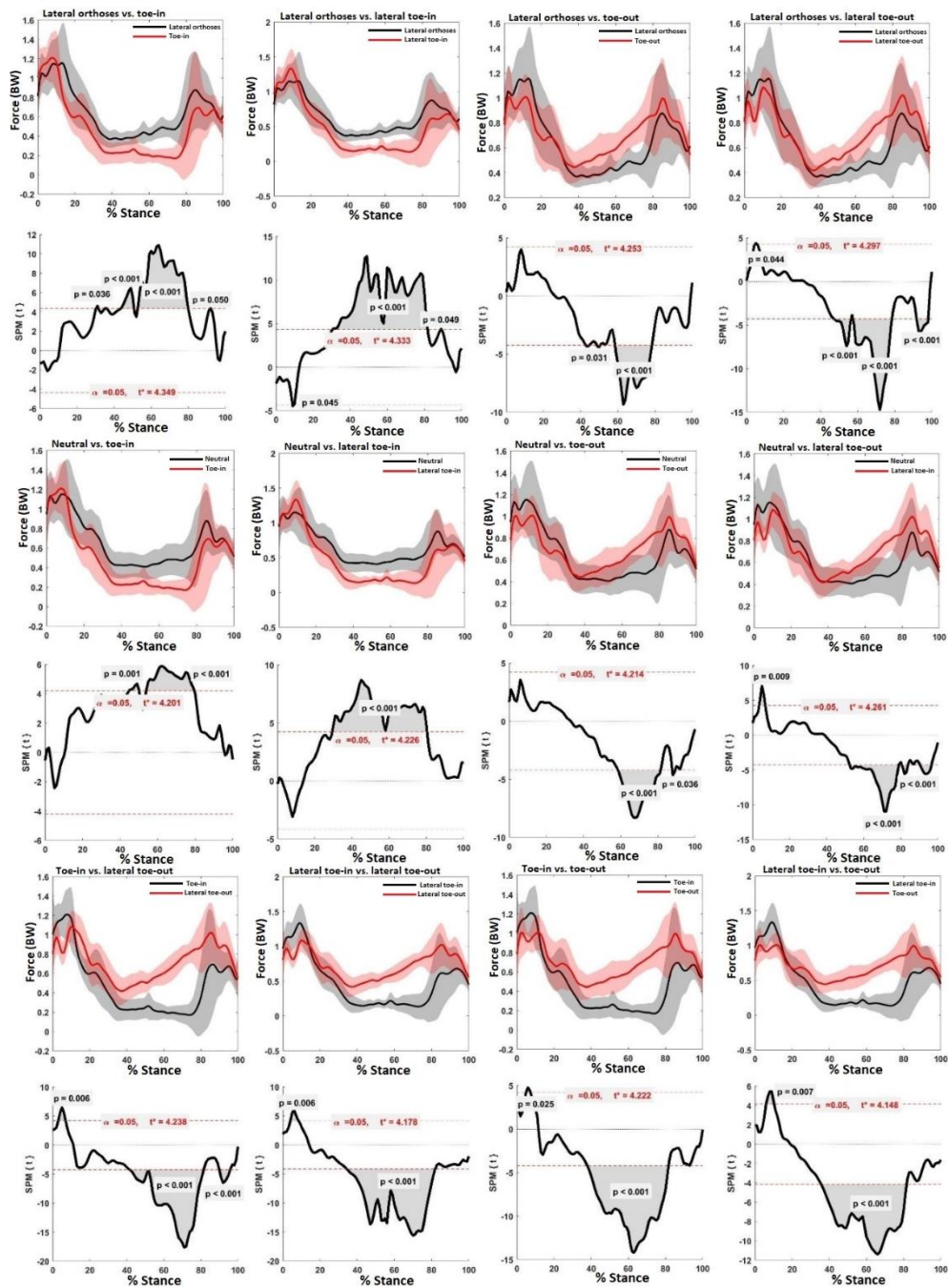
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679 Figure 3. Comparison of medial tibiofemoral forces across the stance phase, in different
 680 conditions. Positive SPM values indicate that values in the first above named condition exceed
 681 those in the other; SPM (t) denotes the t value and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

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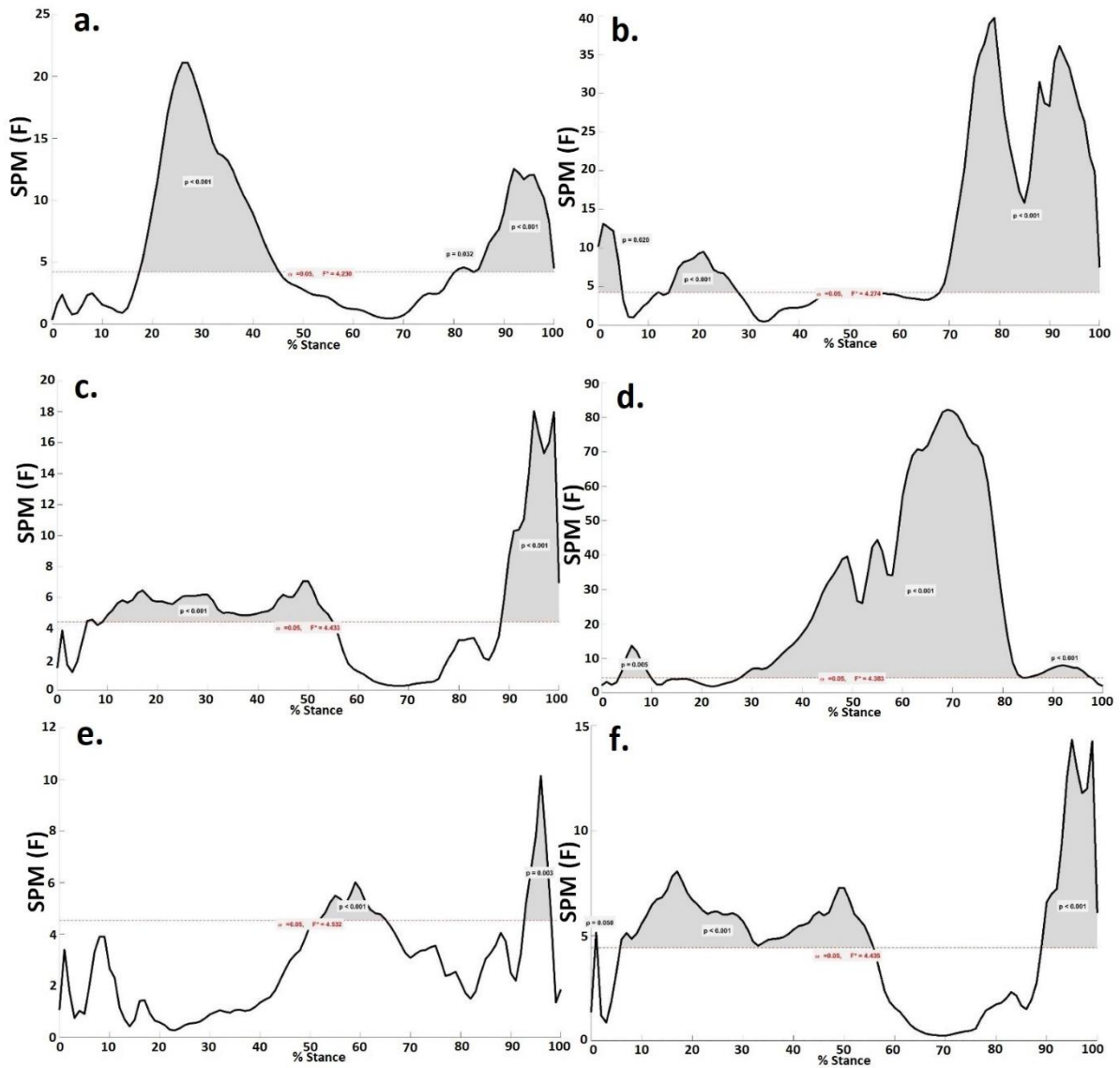


684

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

689

690 **List of supplemental figures**



691

692 1. Comparison of the a. hip joint force, b. medial tibiofemoral force, c. patellofemoral
 693 force, d. lateral tibiofemoral force, e. ankle force and f. patellofemoral stress across the
 694 stance phase in all conditions. SPM (F) denotes the F value, and critical thresholds for
 695 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			