1 **<u>Acute effects of different orthoses on lower extremity kinetics and kinematics during</u>**

2 **<u>running;</u> <u>a musculoskeletal simulation analysis.</u>**

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9 **Abstract**

10 PURPOSE: The current investigation aimed to examine the effects of different orthotic

11 conditions on the biomechanical mechanisms linked to the aetiology of chronic pathologies

12 using musculoskeletal simulation. METHODS: 16 male and 20 females ran over an

13 embedded force plate at 4.0 m/s, in five different conditions (medial, lateral, no-orthoses,

14 semi-custom and off the shelf). Kinematics of the lower extremities were collected using an

15 eight-camera motion capture system and lower extremity joint loading also explored using a

16 musculoskeletal simulation approach. Differences between orthoses conditions were

17 examined using 2 x 2 mixed ANOVA. RESULTS: External instantaneous load rate was

18 significantly reduced in the off the shelf orthoses (male=1290.60 and female=1567.10N/kg/s)

19 compared to the medial (male=1480.45 and female = 1767.05N/kg/s) and semi-custom

20 (male=1552.99 and female=1704.37N/kg/s) conditions. In addition, peak patellofemoral

21 stress was significantly lower in the off the shelf orthoses (male=68.55 and

22 female=94.91KPa/kg) compared to the lateral condition (male=70.49 and

23 female=103.22KPa/kg). Finally, peak eversion angles were significantly attenuated in the

24   medial orthoses (male=-6.61 and female=-7.72deg) compared to the lateral (male=-9.61 and

25   female=-10.32deg), no-orthoses (male=-8.22 and female=-10.10deg), semi-custom (male=-

26   8.25 and female=-9.49deg) and off the shelf (male=-7.54 and female=-8.85deg) conditions.

27   CONCLUSIONS: The current investigation shows that different orthotic devices/

28   configurations may provide distinct benefits in terms of their effectiveness in attenuating the

29   biomechanical parameters linked to the aetiology of chronic running injuries.

30

31   Introduction

32   Regular engagement with distance running has long been associated with a plethora of

33   physiological and psychological advantages. However, due to its cyclical nature, distance

34   running is also associated with an extremely high incidence of chronic pathologies; with an

35   occurrence rate of up to 70 % (Taunton et al., 2002). Specifically, patellofemoral pain, tibial

36   stress fractures, medial tibial stress syndrome, Achilles tendinopathy and pain secondary to

37   hip and knee osteoarthritis are common complaints reported by runners (Taunton et al., 2002,

38   Van Ginckel et al., 2009; Lopes et al., 2012; Snyder et al., 2006).

39

40   Patellofemoral pain is the most common chronic pathology in runners (Taunton et al., 2002).

41   Elevated patellofemoral joint stress is the biomechanical parameter most strongly linked to

42   the aetiology of patellofemoral pain syndrome (Farrokhi et al., 2011). Patellofemoral pain

43   symptoms persist for many years, and importantly >45% of individuals with patellofemoral

44   pain later present with osteoarthritis at this joint (Hinman et al., 2014). In addition,

45   degenerative tibiofemoral joint pathologies account for up to 16.8% of knee pathologies in

46   runners (Taunton et al., 2002). The medial tibiofemoral compartment is considered

47   significantly more prone to degeneration than the lateral aspect (Wise et al., 2012), and the

48   biomechanical parameter most strongly associated with the initiation of knee osteoarthritis is

49   the magnitude of the compressive load experienced at the joint (Morgenroth et al., 2014).

50

51   Furthermore, Achilles tendinopathy is also a common chronic pathology in runners,

52   responsible for up to 15% of all reported injuries (Van Ginckel et al., 2009). Although

53   regarded as the strongest tendon in the body, the Achilles tendon is subjected to forces up to 7

54   * bodyweight during running (Almondroeder et al., 2013). Excessive cyclic stresses borne the

55   tendon are regarded as the main biomechanical stimulus for the initiation of Achilles

56   tendinopathy (Abate et al., 2009). Additionally, medial tibial stress syndrome is similarly a

57   frequently reported chronic running injury cause of running-related injury, accounting for

58   ≥13.6% of all injuries and causing discomfort at the posterio-medial aspect of the tibia

59   (Lopes et al., 2012). The biomechanical mechanisms most prominently linked to the

60   aetiology of medial tibial stress syndrome are the magnitudes of plantarflexion range of

61   motion and hip external rotation range of motion (Hamstra-Wright et al., 2015). Finally, tibial

62   stress fractures are also a serious chronic musculoskeletal injury in runners, representing

63   between 0.5-21.1% of all pathologies (Snyder et al., 2006). The distal-anterior aspect of the

64   tibia is the most frequent location for stress fractures, and retrospective analyses indicate that

65   excessive tibial accelerations/ vertical rates of loading are the biomechanical mechanisms

66   predominantly responsible for the development of stress fractures (Warden et al., 2006).

67

68   Taking into account the rate of chronic pathologies in runners, conservative prophylactic

69   strategies are a key priority for clinical analyses. Foot orthoses are commonly utilized for the

70   prevention/ treatment of chronic running injuries, and a range of foot orthoses are available,

typically classified either as off-the-shelf or custom devices. Off-the-shelf devices are prefabricated by the manufacturer and the design/ fit of the devices are predetermined. Custom orthoses conversely allow the shape, design and fit of the orthotic to be specifically tailored to the individual. However, custom orthoses are typically very expensive and can take several weeks to manufacture. Therefore, orthotic manufacturers have introduced semi-custom devices which can be heat moulded to fit each runner's feet more readily, but at a much lower cost in relation to fully customized devices. In addition to traditional foot orthoses, wedged orthoses that are built up along either the medial or lateral edges have also become common in recent years (Aminian et al., 2014). Wedged devices focus more specifically on modifying the alignment of the lower extremities rather than providing cushioning (Sinclair et al., 2019). Previous clinical analyses have shown that orthoses may be effective in reducing the incidence of lower limb injuries. Bonanno et al., (2018) showed that prefabricated foot orthoses mediated a 34% reduction in the risk of developing medial tibial stress syndrome, patellofemoral pain, Achilles tendinopathy or plantar fasciitis in Australian navy recruits. Similarly, Franklyn-Miller et al., (2011) showed that military officer trainees who received custom orthoses had a significantly reduced absolute injury risk (1 injury per 4666 hours of training) compared to a control group (1 injury per 1600 hours of training). Finally, Sinclair et al., (2018) showed that semi-custom foot orthoses mediated significant reductions patellofemoral pain symptoms in runners from both the strong and weak & tight subgroups of patellofemoral pain patients.

The effects of foot orthoses on lower extremity kinetics and kinematics during running has been explored previously in biomechanical literature. Laughton et al., (2003) and Mündermann et al., (2003) found that off the shelf orthoses significantly reduced tibial accelerations and loading rates during running, although Butler et al., (2003) showed that

96   custom devices had no effect on impact loading parameters. Sinclair et al., (2017) showed

97   that medial orthoses reduced peak eversion and tibial internal rotation, yet Almonroeder et

98   al., (2016) showed using off the shelf devices that eversion/ tibial internal rotation parameters

99   were not significantly affected. In addition, Sinclair et al., (2014) also showed that off the

100 shelf orthoses significantly reduced peak Achilles tendon force, but Sinclair et al., (2015)

101 revealed that semi-custom orthoses had no effect on Achilles tendon kinetics in female

102 runners. Finally, Sinclair, (2018) showed that both medial and lateral orthoses significantly

103 increased patellofemoral kinetics during the stance phase. Foot orthoses are utilized as

104 blanket term for a range of distinct devices that may include off the shelf, custom orthoses,

105 semi-custom devices, heel-lifts, lateral/medial wedges and flat insoles. To date there has yet

106 to be a published investigation of the biomechanical effects of off the shelf, semi-custom, and

107 medial/ lateral orthoses on lower extremity kinetics and kinematics linked to the aetiology of

108 chronic running injuries.

109

110 In addition, previous analyses examining the biomechanical effects of foot orthoses, have

111 utilized joint torque driven musculoskeletal modelling approaches to quantify the loads

112 experienced by the lower extremities. However, as skeletal muscle forces are the main

113 contributors to lower extremity joint loading; musculoskeletal modelling methodologies may

114 not necessarily characterize localized joint kinetics (Herzog et al., 2003). Therefore, more

115 contemporary musculoskeletal simulation based approaches, which allow skeletal muscle

116 forces to be simulated during human movement, and employed as inputs to calculate lower

117 extremity joint reaction forces may be more appropriate (Delp et al., 2007). Such approaches

118 have not yet been adopted to explore biomechanical differences between different orthoses

119 during running.

120

121   Therefore, the aim of the current investigation was to examine the effects of the

122   aforementioned orthotic conditions on the biomechanical mechanisms linked to the aetiology

123   of chronic pathologies, using a musculoskeletal simulation based analysis. An investigation

124   of this nature may provide insight into the potential efficacy of different foot orthoses for the

125   prevention chronic running pathologies.

126

127   **Methods**

128   *Participants*

129   Thirty-six participants (16 male and 20 female) volunteered to take part in the current

130   investigation. The mean and standard deviation characteristics of the participants were (male:

131   age 28.69 ± 6.06 years, height 177.75 ± 5.02 cm, body mass 76.58 ± 8.68 kg and foot posture

132   index $= 3.00 \pm 1.63$ and female: age 32.25 ± 7.36 years, height 161.29 ± 5.61 cm, body mass

133   65.51 ± 7.34 kg and and foot posture index $= 3.90 \pm 2.43$). All identified as recreational

134   runners who trained 3 times/week, completing a minimum of 35 km. Participants were all

135   injury free at the time of data collection and had not undergone lower extremity

136   musculoskeletal surgery. The procedure utilized for this investigation was approved by the

137   University of Central Lancashire, Science, Technology, Engineering and Mathematics,

138   ethical committee (Ref: 874) and all participants provided written informed consent.

139

140   *Orthoses*

Five experimental conditions were examined in this investigation (lateral, medial, semi-custom, off the shelf and no orthotic). For the medial and lateral orthoses, commercially available full-length orthoses (Slimflex Simple, High Density, Full Length, Algeos UK) were examined. The orthoses were able to be modified to either a 5˚ varus or valgus configuration which in two separate components spanning the full length of the device. 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 semi-custom insoles (Sole Control, Sole, Milton Keynes, UK), were made from ethylene-vinyl acetate with a shore A 30 hardness rating and a heel thickness of 6 mm. To mould the insoles, they were placed into a pre-heated oven (90 °C) for a duration of two minutes. The heated insoles were then placed inside the shoes and participants were asked to stand upright without moving for two minutes to allow the process of moulding the insoles to the longitudinal arch profile of each participant, in accordance with manufacturer instructions. The off the shelf orthoses (Sorbothane, shock stopper sorbo Pro, Nottinghamshire, UK) were made from a custom polyurethane polymer and had a heel thickness of 6 mm and a shore A hardness rating of 10. To ensure consistency each participant wore the same footwear (Asics, Patriot 6). The experimental footwear had a mean mass of 0.265 kg, heel thickness of 22 mm and heel drop of 10 mm. The order that participants ran in each orthotic condition was counterbalanced.

Procedure

Participants ran across a 20 m biomechanics laboratory surface (MondoSport Ramflex, Mondo, Italy) at 4.0 m/s (± 5%), striking an embedded piezoelectric force platform (Kistler, Kistler Instruments Ltd., Alton, Hampshire), which sampled at 1000 Hz, with their right (dominant) foot. Running velocity was monitored using infrared timing gates (Newtest, Oy

165   Koulukatu, Finland). The stance phase was delineated as the duration over which 20 N or

166   greater of vertical force was applied to the force platform. Runners completed five successful

167   trials in each of the five different orthotic conditions. A successful trial was defined as one

168   within the specified velocity range, where all tracking clusters were in view of the cameras,

169   the foot made full contact with the force plate and there was no evidence of gait

170   modifications due to the experimental conditions. The order that participants ran in each

171   condition was counterbalanced, by providing each orthotic with a letter from A-E and block

172   counterbalancing the order in which each was presented to each participant. Kinematics and

173   ground reaction forces data were synchronously collected. Kinematic data was captured at

174   250 Hz via an eight-camera motion analysis system (Qualisys Medical AB, Goteburg,

175   Sweden). Dynamic calibration of the motion capture system was performed before each data

176   collection session.

177

178   After being tested in each orthotic condition, participants were asked to provide their rating

179   of the comfort of each one. The comfort measurement procedure consisted of a 150 mm

180   visual analogue scale with the extreme left side being indicative of 'not comfortable at all'

181   and the extreme right of the scale labelled as 'most comfortable condition imaginable'

182   (Mündermann et al., 2003). Upon conclusion of the data collection, participants were also

183   asked to subjectively indicate which orthotic condition that they preferred.

184

185   To define the anatomical frames of the thorax, pelvis, thighs, shanks and feet retroreflective

186   markers were placed at the C7, T12 and xiphoid process landmarks and also positioned

187   bilaterally onto the acromion process, iliac crest, anterior superior iliac spine (ASIS),

188   posterior super iliac spine (PSIS), medial and lateral malleoli, medial and lateral femoral

189   epicondyles, greater trochanter, calcaneus, first metatarsal and fifth metatarsal. Carbon-fibre

190   tracking clusters comprising of four non-linear retroreflective markers were positioned onto

191   the thigh and shank segments. In addition to these, the foot segments were tracked via the

192   calcaneus, first metatarsal and fifth metatarsal, the pelvic segment was tracked using the PSIS

193   and ASIS markers and the thorax segment was tracked using the T12, C7 and xiphoid

194   markers. Static calibration trials were obtained with the participant in the anatomical position

195   in order for the positions of the anatomical markers to be referenced in relation to the tracking

196   clusters/markers. A static trial was conducted with the participant in the anatomical position

197   in order for the anatomical positions to be referenced in relation to the tracking markers,

198   following which those not required for dynamic data were removed.

199

200   To measure axially directed accelerations at the tibia, an accelerometer (Biometrics ACL 300,

201   Gwent United Kingdom) sampling at 1000Hz was used. The device was mounted onto a

202   piece of lightweight carbon-fibre material using the protocol outlined by Sinclair et al.,

203   (2013). The accelerometer was attached securely to the distal anterio-medial aspect of the

204   tibia in alignment with its longitudinal axis, 0.08 m above the medial malleolus. Strong non-

205   stretch adhesive tape was placed over the device and leg to avoid overestimating the

206   acceleration due to tissue artefact (Sinclair et al., 2013).

207

208   The Achilles tendon of each participant's examined (right) side was inspected using

209   ultrasound imaging (SonoScope A6, Sonomed, China). Each participant laid face downwards

210   on a physiotherapy table with their ankle joint in a neutral position. A 46 mm 5-11 MHz

211   linear ultrasound probe (model L745) was placed perpendicular to the Achilles tendon,

212   between the medial and lateral malleoli (Milgrom et al., 2014). The medial-lateral and

213 anterior-posterior dimensions were recorded, and the cross-sectional area was calculated

214 using the associated formula for an oval i.e. Anterior-posterior * medial-lateral * π / 4

215 (Milgrom et al., 2014). Three images were obtained from each participant and the mean of

216 these recordings was calculated.

217

218 *Processing*

219 Dynamic trials were digitized using Qualisys Track Manager in order to identify anatomical

220 and tracking markers, then exported as C3D files to Visual 3D (C-Motion, Germantown, MD,

221 USA). All data were normalized to 100% of the stance phase then processed trials were

222 averaged within subjects for statistical analysis. Ground reaction force and kinematic data

223 were smoothed using cut-off frequencies of 50 and 12 Hz with a low-pass Butterworth 4th

224 order zero lag filter (Sinclair, 2018). All net force parameters throughout were normalized by

225 dividing by body mass (N/kg). Three-dimensional kinematic measures were extracted using

226 Visual 3D from the hip, knee, ankle that were extracted for statistical analysis were 1) angle

227 at footstrike, 2) peak angle during the stance phase and 3) angular range of motion (ROM)

228 from footstrike to peak angle. In addition, tibial internal rotation kinematics were also

229 calculated in accordance with Eslami et al., (2007). From the force platform, the external

230 instantaneous loading rate (N/kg/s) was calculated by obtaining the peak increase in force

231 between adjacent data points. In addition, the tibial acceleration signal was filtered using a 60

232 Hz Butterworth zero lag 4th order low pass filter (Sinclair et al., 2013), and the peak tibial

233 acceleration (g) was extracted as the highest positive acceleration peak during the stance

234 phase.

235

236 Data during the stance phase were exported from Visual 3D into OpenSim 3.3 software

237 (Simtk.org). A validated musculoskeletal model with 12 segments, 19 degrees of freedom

238 and 92 musculotendon actuators (Lerner et al., 2015) was used to estimate extremity joint

239 forces. The model was scaled for each participant to account for the anthropometrics of each

240 athlete. As muscle forces are the main determinant of joint compressive forces (Herzog et al.,

241 2003), muscle kinetics were quantified using static optimization. Peak compressive

242 patellofemoral, medial/ lateral tibiofemoral, ankle and hip joint forces were calculated via the

243 joint reaction analyses function using the muscle forces generated from the static

244 optimization process. Furthermore, peak patellofemoral stress (KPa/kg) was quantified by

245 dividing the patellofemoral force by the contact area. Patellofemoral contact areas were

246 obtained by fitting a polynomial curve to the sex specific data of Besier et al., (2005), who

247 estimated patellofemoral contact areas as a function of the knee flexion angle using MRI.

248

249 Achilles tendon forces were estimated in accordance with the protocol of Almonroeder et al.,

250 (2013), by summing the muscle forces of the medial gastrocnemius, lateral, gastrocnemius,

251 and soleus muscles. In addition, Achilles tendon stress was estimated by dividing the Achilles

252 tendon forces by the cross-sectional area of the tendon measured from the ultrasound images.

253 Peak Achilles tendon force (N/kg) and stress (KPa/kg) were extracted for statistical analysis.

254

255 In addition, patellofemoral, medial/ lateral tibiofemoral, ankle, hip and Achilles tendon

256 instantaneous load rates (N/kg/s and KPa/kg/s) were also extracted by obtaining the peak

257 increase in force/ stress between adjacent data points. Finally, the integral of the hip,

258 tibiofemoral, ankle, patellofemoral and Achilles tendon forces (N/kg·s) and stresses

259 (KPa/kg·s) during the stance phase were calculated using a trapezoidal function.

260

261 *Statistical analyses*

262 Descriptive statistics of means and standard deviations were obtained for each outcome

263 measure and for each orthotic condition. Shapiro-Wilk tests were used to screen the data for

264 normality. Differences in biomechanical parameters were examined using 5 (ORTHOTIC) x

265 2 (GENDER) mixed ANOVA's and differences in comfort ratings were examined using 4

266 (ORTHOTIC) x 2 (GENDER) mixed ANOVA's. Statistical significance was accepted at the

267 $P \leq 0.05$ level and effect sizes for all significant findings were calculated using partial Eta^2

268 ($p\eta^2$). In the event of a significant main effect, pairwise comparisons were performed.

269 Finally, a chi-squared (χ^2) test was utilised to test the assumption that an equal number of

270 participants would subjectively favour each of the orthotic conditions. All statistical actions

271 were conducted using SPSS v25.0 (SPSS Inc, Chicago, USA).

272

273 **Results**

274 *Joint kinetics*

275 Medial tibiofemoral joint

276 At the medial aspect of the tibiofemoral joint, there was a main effect of GENDER ($P<0.05$,

277 $p\eta^2=0.34$) for the peak medial tibiofemoral force, with peak force being greater in male

278 runners. In addition, there was a main effect of GENDER ($P<0.05$, $p\eta^2=0.33$) for the medial

279 tibiofemoral integral, with the medial tibiofemoral integral being greater in males.

280

281 Lateral tibiofemoral joint

282   At the lateral aspect of the tibiofemoral joint, there was a main effect of GENDER (P<0.05,

283    $p\eta^2 = 0.38$) for the peak lateral tibiofemoral force, with peak force being greater in male

284   runners. In addition, there was a main effect of ORTHOTIC (P<0.05, $p\eta^2=0.38$). Post-hoc

285   pairwise comparisons showed that the peak lateral tibiofemoral force was significantly

286   greater in the lateral (P=0.023) condition, compared to the medial orthoses. In addition, there

287   was a main effect of GENDER (P<0.05, $p\eta^2=0.16$) for the lateral tibiofemoral instantaneous

288   loading rate, with this parameter being greater in male runners. In addition, there was a main

289   effect of ORTHOTIC (P<0.05, $p\eta^2=0.10$). Post-hoc pairwise comparisons showed that the

290   lateral tibiofemoral instantaneous loading rate was significantly greater in the lateral

291   (P=0.025) condition, compared to the medial orthoses. Finally, there was a main effect of

292   GENDER (P<0.05, $p\eta^2=0.35$) for the lateral tibiofemoral force integral, with this value being

293   greater in male runners.

294

295   @@@TABLE 1 NEAR HERE@@@

296

297   Patellofemoral joint

298   A main effect of ORTHOTIC (P<0.05, $p\eta^2=0.09$) was found for peak patellofemoral force.

299   Post-hoc pairwise comparisons showed that peak patellofemoral force was significantly

300   larger in the lateral condition (P=0.039) compared to the off the shelf orthoses. For peak

301   patellofemoral stress there was a main effect of ORTHOTIC (P<0.05, $p\eta^2=0.09$). Post-hoc

302   pairwise comparisons showed that peak patellofemoral stress was significantly larger in the

303   lateral condition (P=0.04) compared to the off the shelf orthoses. In addition, there was also a

304   main effect of GENDER (P<0.05, $p\eta^2=0.35$), with peak stress being greater in females. For

305 the patellofemoral stress instantaneous loading rate, a main effect of GENDER (P<0.05,

306 pη^2=0.25) was found, with this parameter being greater in females. For the patellofemoral

307 force integral a main effect of ORTHOTIC (P<0.05, pη^2=0.10) was found. Post-hoc pairwise

308 comparisons showed that patellofemoral force integral was significantly larger in the lateral

309 condition, compared to no orthotic (P=0.04) off the shelf orthoses (P=0.018). There was also

310 a main effect of ORTHOTIC (P<0.05, pη^2=0.09) for the patellofemoral stress integral. Post-

311 hoc pairwise comparisons showed that the patellofemoral stress integral was significantly

312 larger in the lateral condition (P=0.015), compared to the off the shelf orthoses. In addition,

313 there was also a main effect of GENDER (P<0.05, pη^2=0.37), the patellofemoral stress

314 integral being greater in females.

315

316 Ankle joint

317 At the ankle, there was a main effect of GENDER (P<0.05, pη^2=0.36) for the peak ankle

318 force, with this measurement being larger in males. For the integral of the ankle force

319 (P<0.05, pη^2=0.24), a main effect of GENDER was found, with the ankle force integral being

320 larger in males.

321

322 **@@@TABLE 2 NEAR HERE@@@**

323

324 *Achilles tendon kinetics*

325 There was a main effect of GENDER for both the peak Achilles tendon force (P<0.05,

326 pη^2=0.41) and stress (P<0.05, pη^2=0.40), with both parameters being greater in male runners.

327   In addition, there was a main effect of GENDER for both the Achilles tendon force ($P<0.05$,

328    $p\eta^2=0.36$) and stress ($P<0.05$, $p\eta^2=0.35$) instantaneous loading rates, with both parameters

329   being greater in male runners. Finally, for the integral of the Achilles tendon force ($P<0.05$,

330    $p\eta^2=0.18$) and stress ($P<0.05$, $p\eta^2=0.19$), a main effect of GENDER was found, with both

331   measures being larger in males.

332

333   *External instantaneous loading rate and tibial accelerations*

334   For the external instantaneous loading rate, there was a main effect for ORTHOTIC ($P<0.05$,

335    $p\eta^2=0.10$). Post-hoc pairwise comparisons showed that the instantaneous loading rate was

336   significantly greater in the medial ($P=0.028$) and semi-custom ($P=0.03$) conditions compared

337   to the off the shelf orthoses. For peak tibial acceleration, there was a main effect for

338   ORTHOTIC ($P<0.05$, $p\eta^2=0.11$). Post-hoc pairwise comparisons showed that the peak tibial

339   accelerations were significantly greater in the semi-custom ($P<0.001$) conditions compared to

340   the off the shelf orthoses. In addition, there was also a main effect of GENDER ($P<0.05$, $p\eta^2$

341    $= 0.13$), with tibial accelerations being greater in females.

342

343   *Subjective ratings*

344   There was a main effect of ORTHOTIC ($P<0.05$, $p\eta^2=0.51$) for participants ratings of

345   comfort. Post-hoc pairwise comparisons showed that the semi-custom ($P<0.001$ & $P<0.001$)

346   and off the shelf ($P<0.001$ & $P<0.001$) orthoses were rated as being significantly more

347   comfortable than the medial and lateral conditions. Finally, the semi-custom orthoses were

348   rated as being significantly ($P=0.029$) more comfortable than the off the shelf condition. In

349   addition, the Chi-squared analysis of orthotic preferences was significant ($\chi^2_{(3)}=22.00$,

350 P<0.05) with 19 participants selecting the semi-custom orthoses, 12 off the shelf, 4 medial

351 and 1 the lateral conditions.

352

353

354

355 *Joint kinematics*

356 Hip

357 For the peak hip adduction angle there was a main effect of ORTHOTIC (P<0.05, $p\eta^2$=0.20).

358 Post-hoc pairwise comparisons showed that peak adduction was significantly greater in the

359 lateral and semi-custom orthoses compared to the medial (P<0.001 & P=0.002), no orthotic

360 (P=0.002 & P=0.036) and off the shelf orthoses (P<0.001 & P<0.001). There was also a main

361 effect of GENDER (P<0.05, $p\eta^2$=0.14), with peak adduction being larger in females.

362

363 Knee

364 For the sagittal knee angle at footstrike there was a main effect of GENDER (P<0.05,

365 $p\eta^2$=0.18), with knee flexion being larger in females. There was also a main effect of

366 GENDER (P<0.05, $p\eta^2$=0.20) for the peak knee flexion angle, which was shown to be greater

367 in females. There was also a main effect of ORTHOTIC (P<0.05, $p\eta^2$=0.11) for the peak

368 knee abduction angle. Post-hoc pairwise comparisons showed that peak abduction was

369 significantly larger in the lateral (P=0.032) and semi-custom orthoses (P=0.01) compared to

370 the no orthotic condition.

371

372

373 For the sagittal ankle angle at footstrike there was a main effect of GENDER (P<0.05,

374 $p\eta^2$=0.25), with dorsiflexion being larger in females. In addition, there was also a main effect

375 of ORTHOTIC (P<0.05, $p\eta^2$=0.13) for the peak dorsiflexion angle. Post-hoc pairwise

376 comparisons showed that peak dorsiflexion was significantly greater in the medial orthoses

377 compared to the lateral (P=0.04), no orthotic (P=0.028), off the shelf (P=0.012) and semi-

378 custom (P=0.01) conditions. There was also a main effect of GENDER (P<0.05, $p\eta^2$= 0.22)

379 for dorsiflexion ROM, with this measurement being larger in males.

380

381 For the peak eversion angle there was a main effect of ORTHOTIC (P<0.05, $p\eta^2$=0.26). Post-

382 hoc pairwise comparisons showed that peak eversion was significantly greater in the lateral

383 (P<0.001), no orthotic (P<0.001), off the shelf (P<0.032) and semi-custom (P<0.001)

384 conditions compared to medial orthoses. In addition, for the eversion ROM there was a main

385 effect of ORTHOTIC (P<0.05, $p\eta^2$=0.61). Post-hoc pairwise comparisons showed that

386 eversion ROM was significantly greater in the lateral (P<0.001), no orthotic (P<0.001), off

387 the shelf (P<0.001) and semi-custom (P<0.001) conditions compared to the medial orthoses.

388 In addition, peak eversion was significantly larger in the lateral orthoses compared to the off

389 the shelf (P<0.001), semi-custom (P<0.001) and no orthotic (P=0.005) conditions.

390

391

392 For the peak tibial internal rotation angle there was a main effect of ORTHOTIC (P<0.05,

393 $p\eta^2$=0.28). Post-hoc pairwise comparisons showed that peak tibial internal rotation was

394 significantly greater in the lateral orthoses compared to the medial (P<0.001) no orthotic

395 (P<0.001), off the shelf (P<0.001) and semi-custom (P<0.017) conditions. In addition, peak

396 tibial internal rotation was significantly greater in the semi-custom orthoses compared to the

397 medial (P<0.001) and off the shelf (P=0.001) conditions. In addition, for the tibial internal

398 rotation ROM there was a main effect of ORTHOTIC (P<0.05, $p\eta^2$=0.30). Post-hoc pairwise

399 comparisons showed that tibial internal rotation ROM was significantly greater in the lateral

400 (P<0.001), no orthotic (P<0.001), off the shelf (P=0.001) and semi-custom (P<0.001)

401 conditions compared to the medial orthoses. In addition, tibial internal rotation ROM was

402 also significantly greater in the lateral (P=0.04), no orthotic (P=0.027) and semi-custom

403 orthoses (P=0.001) compared to the off the shelf condition.

404

405   **@@@TABLE 4 NEAR HERE@@@**

406   **@@@TABLE 5 NEAR HERE@@@**

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408

409 **Discussion**

410 This study aimed to examine the effects of different orthotic conditions on the biomechanical

411 mechanisms linked to the aetiology of chronic pathologies. To the authors knowledge this is

412 the first investigation to collectively explore the effects of different orthoses on lower

413 extremity kinetics and kinematics during running, and may provide insight into the potential

414 efficacy of different foot orthoses for the prevention chronic running pathologies.

415

416   Patellofemoral pain is regarded as the most common chronic running injury (Taunton et al.,

417   2002). Females are renowned for being at increased risk from patellofemoral disorders;

418   therefore, it is important that the current investigation showed female runners to be associated

419   with increased patellofemoral loading. This observation concurs with those of Sinclair &

420   Selfe, (2015) and given the proposed relationship between joint stress and patellofemoral

421   pathology (Farrokhi et al., 2011), appears to provide insight into the responsible factors for

422   the increased incidence of patellofemoral pain in females. In support of the findings of

423   Sinclair, (2018), the current investigation also showed that patellofemoral joint stress

424   parameters were significantly greater when running in the lateral orthoses in relation to

425   running in off the shelf devices. Although the mean difference between these orthotic

426   conditions was relatively small, this observation may nonetheless be clinically important, as

427   patellofemoral pain symptoms are believed to be initiated via excessive/ repeated

428   patellofemoral joint stress (Farrokhi et al., 2011). The current study indicates that running

429   with off the shelf orthoses may be preferable over lateral wedged devices, as a mechanism to

430   reduce the risk from the biomechanical parameters linked to the aetiology of patellofemoral

431   pain in runners.

432

433   At the tibiofemoral joint, there was no effect of orthoses at the medial aspect. This opposes

434   previous walking analyses, which have consistently shown that lateral orthoses reduce the

435   magnitude of the external knee adduction moment (Jones et al., 2013). It is proposed that the

436   difference between analyses relates to the manner in which tibiofemoral loading was

437   calculated in the current study, as previous analyses have used coronal plane joint torques as

438   a pseudo measure of medial compartment loading, which do not account for muscular co-

439   contraction about the knee joint (Herzog et al., 2003). However, at the lateral aspect of the

440   tibiofemoral joint compressive loading was significantly greater in the lateral orthoses in

441 relation to the medial devices. This indicates that although lateral orthoses were not able to

442 attenuate compressive loading at the medial aspect of the joint, they were able to transfer load

443 to the lateral tibiofemoral compartment. Therefore, although the increases in compressive

444 load were small, lateral wedged devices may place runners at greater risk from the

445 mechanisms associated with tibiofemoral pathologies. Furthermore, in contrast, to the

446 findings at the patellofemoral joint, this investigation showed that at both the medial and

447 lateral aspects of the tibiofemoral joint males were associated with statistically greater joint

448 loading parameters in relation to females. Leading to the conclusion that males are at greater

449 risk from the biomechanical parameters linked to the aetiology of tibiofemoral pathologies.

450

451 In agreement with the findings of Greenhalgh & Sinclair, (2014) the current study also

452 showed that males were associated with increased Achilles tendon stress and ankle joint force

453 parameters. In contrast to patellofemoral pathologies, males are at increased risk from

454 Achilles tendinopathies in relation to age-matched females (Hess, 2010). Given the proposed

455 association between tendon stress and the physiological initiation of tendinous collagen

456 degradation (Abate et al., 2009), this observation appears to provide further insight into the

457 biomechanical mechanisms behind the increased incidence Achilles tendinopathy in males.

458 However, as there were no significant differences between orthoses in ankle or Achilles

459 tendon load parameters, the observations from this investigation are in contrast to those of

460 Sinclair et al., (2014) who showed that off the shelf orthoses significantly reduced peak

461 Achilles tendon force, but agree with those of Sinclair et al., (2015) with regards to semi-

462 custom devices. As such, the findings from this study using musculoskeletal simulation

463 indicate that foot orthoses do not influence the biomechanical parameters linked to the

464 aetiology of ankle/ Achilles tendon pathologies during running.

465

466   Importantly, in agreement with the findings of Mündermann et al., (2003) and Sinclair et al.,

467   (2014), this study also showed that instantaneous loading rates and peak tibial accelerations

468   were significantly larger in the medial and semi-custom conditions compared to off the shelf

469   orthoses. Excessive tibial accelerations/ vertical rates of loading are the biomechanical

470   mechanisms responsible for the development of stress fractures (Warden et al., 2006).

471   Therefore, this study indicates that off the shelf orthoses may be effective in attenuating the

472   mechanisms linked to the aetiology of tibial stress fractures in runners. In addition, that

473   females were associated with increased tibial accelerations may also be clinically important

474   taking into account their proposed link to the aetiology of stress fractures and may provide

475   further insight into the biomechanical mechanisms responsible for the increased incidence of

476   stress fractures in female runners (Jones et al., 1993).

477

478   In conclusion, although the biomechanical effects of foot orthoses have been examined

479   previously, current knowledge with regards to the effects of different orthoses is limited. This

480   study therefore adds to the current literature by examining the influence of different orthoses

481   on the biomechanical mechanisms linked to the aetiology of chronic pathologies, using

482   musculoskeletal simulation. The current investigation importantly showed that patellofemoral

483   stress parameters and loading rates/ peak tibial accelerations were significantly reduced in the

484   off the shelf orthoses and lateral tibiofemoral loading parameters were significantly

485   attenuated in the medial orthotic condition. Therefore, the current investigation indicates that

486   different orthotic devices/ configurations may provide distinct benefits in terms of their

487   effectiveness in attenuating the biomechanical parameters linked to the aetiology of chronic

488   running injuries.

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611 Table 1: Hip and knee joint kinetics (Mean & SD) for each orthotic condition.

	Male										
	Medial		Lateral		No-orthoses		Semi-custom		Off the shelf		
	Mean	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*	
Peak hip force (N/kg)	88.48	7.32	90.33	8.34	89.99	8.29	91.42	10.18	87.97	8.18	
Hip force instantaneous load rate (N/kg/s)	3307.86	913.51	3315.23	669.65	3828.30	786.74	3839.14	1117.64	3589.09	803.50	
Hip integral (N/kg·s)	13.03	1.79	13.55	1.77	13.25	1.88	13.17	2.02	12.95	1.58	
Peak medial tibiofemoral force (N/kg)	71.02	8.45	73.66	9.70	71.79	9.71	71.24	12.16	74.79	9.95	B
Medial tibiofemoral instantaneous load rate (N/kg/s)	2434.42	536.84	2591.40	567.63	2914.90	850.45	2599.01	894.05	2475.01	771.44	
Medial tibiofemoral integral (N/kg·s)	9.03	1.15	9.37	1.23	9.13	1.30	9.09	1.51	9.16	1.07	B
Peak lateral tibiofemoral force (N/kg)	45.44	12.53	48.04	14.86	48.93	14.44	49.37	16.16	48.50	11.15	A, B
Lateral tibiofemoral instantaneous load rate (N/kg/s)	1773.79	583.72	1959.83	679.00	1849.62	598.64	1947.66	690.18	1859.87	466.90	A, B
Lateral tibiofemoral integral (N/kg·s)	4.68	0.85	4.59	1.20	4.67	0.91	4.72	1.12	4.78	0.77	B
	Female										
	Medial		Lateral		No-orthoses		Semi-custom		Off the shelf		
	Mean	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*	
Peak hip force (N/kg)	84.75	8.47	85.26	8.48	85.36	10.78	86.45	9.85	85.43	8.15	
Hip force instantaneous load rate (N/kg/s)	3285.00	882.75	3281.45	799.94	3010.18	588.48	3396.73	1042.38	3387.98	1122.21	
Hip integral (N/kg·s)	11.82	1.60	12.43	1.22	12.03	1.63	12.33	1.58	11.93	1.50	
Peak medial tibiofemoral force (N/kg)	60.20	13.01	58.56	9.76	57.27	13.15	59.59	10.91	58.97	10.95	B
Medial tibiofemoral instantaneous load rate	2529.12	1153.93	2542.29	995.54	2346.42	802.95	2482.47	932.52	2425.37	975.89	

(N/kg/s)											
Medial tibiofemoral integral (N/kg·s)	7.29	1.44	7.49	1.42	7.25	1.75	7.44	1.62	7.25	1.62	B
Peak lateral tibiofemoral force (N/kg)	32.66	7.41	35.54	6.59	34.52	8.38	34.98	7.89	32.50	6.65	A, B
Lateral tibiofemoral instantaneous load rate (N/kg/s)	1428.72	406.22	1616.61	483.48	1523.92	521.47	1578.85	461.88	1374.27	306.65	A, B
Lateral tibiofemoral integral (N/kg·s)	3.51	0.83	3.76	0.86	3.56	0.95	3.62	0.84	3.42	0.72	B

Key: A = main effect of ORTHOSES & B = main effect of GENDER

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629   Table 2: Patellofemoral and joint kinetics (Mean & SD) for each orthotic condition.

	Male										
	Medial		Lateral		No-orthoses		Semi-custom		Off the shelf		
	Mean	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*	
Peak ankle force (N/kg)	115.69	22.41	118.63	15.40	117.87	19.59	121.14	21.30	120.16	15.85	B
Ankle force instantaneous load rate (N/kg/s)	3129.58	1059.63	3218.30	649.58	3334.58	941.19	3227.41	509.45	3148.57	656.10	
Ankle integral (N/kg·s)	13.48	2.61	13.95	1.65	13.75	2.35	14.39	2.67	14.08	2.13	B
Peak patellofemoral force (N/kg)	40.26	14.78	40.54	16.90	39.00	13.16	40.01	14.42	39.14	13.50	A
Peak patellofemoral stress (KPa/kg)	70.56	22.11	70.49	25.69	68.92	19.93	70.15	21.80	68.55	20.63	A, B
Patellofemoral force instantaneous load rate (N/kg/s)	1272.87	339.23	1274.20	339.02	1306.85	380.22	1310.70	336.69	1217.09	268.24	
Patellofemoral stress instantaneous load rate (KPa/kg/s)	2466.63	585.35	2477.26	429.23	2782.31	877.60	2721.66	588.61	2506.96	602.10	B
Patellofemoral force integral (N/kg·s)	3.10	1.31	3.33	1.73	2.95	1.13	3.03	1.35	3.13	1.28	A
Patellofemoral stress integral (KPa/kg·s)	5.60	2.11	5.90	2.83	5.40	1.89	5.50	2.22	5.60	2.08	A, B
	Female										
	Medial		Lateral		No-orthoses		Semi-custom		Off the shelf		
	Mean	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*	
Peak ankle force (N/kg)	96.42	16.52	98.71	12.73	98.83	16.37	97.52	17.63	95.96	14.61	B
Ankle force instantaneous load rate (N/kg/s)	3013.14	736.42	3020.20	631.00	2817.86	679.30	3028.02	681.18	2960.76	789.04	
Ankle integral (N/kg·s)	11.72	2.05	12.05	1.84	11.73	2.25	11.83	2.30	11.62	1.87	B
Peak patellofemoral force (N/kg)	46.86	14.56	48.56	12.39	44.59	10.83	49.01	16.86	44.39	11.53	A
Peak patellofemoral stress (KPa/kg)	100.28	24.13	103.22	20.69	96.57	17.88	104.41	30.19	94.91	18.83	A, B

Patellofemoral force instantaneous load rate (N/kg/s)	1473.54	521.20	1423.69	409.31	1388.64	517.25	1390.18	354.61	1367.60	486.44	
Patellofemoral stress instantaneous load rate (KPa/kg/s)	3785.04	1398.42	3633.07	1118.76	3658.16	1305.26	3667.80	949.96	3584.23	1450.64	B
Patellofemoral force integral (N/kg·s)	4.01	1.43	4.15	1.20	3.89	1.33	4.14	1.74	3.76	1.31	A
Patellofemoral stress integral (KPa/kg·s)	9.00	2.55	9.30	2.03	8.80	2.28	9.30	3.25	8.50	2.32	A, B

Key: A = main effect of ORTHOSES & B = main effect of GENDER

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648   Table 3: Achilles tendon, loading rate and tibial acceleration parameters (Mean & SD) for each
649   orthotic condition.

	Medial		Lateral		No-orthoses		Semi-custom		Off the shelf		
	Mean	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*	
Comfort	5.92	2.91	6.00	3.33			11.83	2.21	10.33	3.20	A
External instantaneous load rate (N/kg/s)	1480.45	525.84	1383.08	356.07	1562.52	431.02	1552.99	419.77	1290.60	395.12	A
Peak tibial acceleration (g)	7.09	2.26	7.35	1.95	7.07	1.88	7.93	1.94	6.91	1.71	A, B
Peak Achilles tendon force (N/kg)	75.54	10.23	75.77	6.75	76.19	14.36	77.77	13.95	78.64	11.56	B
Peak Achilles tendon stress (KPa/kg)	1569.68	212.50	1574.58	140.27	1583.26	298.50	1616.16	289.90	1634.15	240.26	B
Achilles tendon instantaneous load rate (N/kg/s)	1650.18	445.92	1539.91	239.20	1703.98	550.80	1587.40	309.96	1632.10	415.57	B
Achilles tendon stress instantaneous load rate (KPa/kg/s)	34290.99	9266.24	31999.52	4970.71	35408.90	11445.66	32986.31	6440.98	33915.23	8635.62	B
Achilles tendon force integral (N/kg·s)	7.80	1.42	7.94	0.74	7.84	1.88	8.26	1.72	8.19	1.46	B
Achilles tendon stress integral (KPa/kg·s)	162.13	29.53	164.96	15.45	162.93	39.17	171.60	35.68	170.27	30.30	B
	Female										
	Medial		Lateral		No-orthoses		Semi-custom		Off the shelf		
	Mean	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*	
Comfort	5.45	3.91	6.65	3.45			11.95	3.32	10.45	2.87	A
External instantaneous load rate (N/kg/s)	1767.05	950.24	1629.06	600.96	1669.17	648.25	1704.37	526.02	1567.10	712.42	A
Peak tibial acceleration (g)	8.72	2.15	8.90	2.21	8.70	2.42	9.01	2.12	8.55	2.09	A, B
Peak Achilles tendon force (N/kg)	61.53	12.32	61.39	10.86	60.93	11.67	61.96	12.60	60.89	10.26	B
Peak Achilles tendon stress (KPa/kg)	1278.52	255.94	1275.66	225.70	1266.16	242.42	1287.52	261.73	1265.29	213.26	B
Achilles tendon instantaneous load rate (N/kg/s)	1285.07	327.89	1211.65	244.72	1136.86	270.52	1286.43	348.36	1244.78	322.38	B
Achilles tendon stress instantaneous load rate	26703.86	6813.52	25178.27	5085.26	23624.08	5621.48	26732.19	7239.00	25866.78	6699.04	B

(KPa/kg/s)											
Achilles tendon force integral (N/kg·s)	6.81	1.61	6.84	1.40	6.66	1.66	6.82	1.66	6.70	1.34	B
Achilles tendon stress integral (KPa/kg·s)	141.50	33.37	142.06	29.11	138.40	34.40	141.64	34.43	139.33	27.75	B

Key: A = main effect of ORTHOSES & B = main effect of GENDER

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669   Table 4: Three-dimensional hip joint kinematics (Mean & SD) for each orthotic condition.

	Male									
	Medial		**Lateral**		**No-orthoses**		**Semi-custom**		**Off the shelf**	
	Mean	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*
Sagittal plane										
Angle at footstrike (°)	38.15	14.28	39.36	13.40	38.07	14.42	40.77	8.41	37.69	12.84
Peak flexion (°)	38.77	14.08	39.90	13.32	38.34	14.04	41.20	8.26	38.33	12.33
ROM (°)	0.62	1.26	0.54	1.04	0.28	0.60	0.43	0.67	0.63	1.06
Coronal plane										
Angle at footstrike (°)	-0.04	8.99	1.02	8.93	0.07	9.69	1.64	6.75	-0.21	9.05
Peak adduction (°)	7.77	8.57	9.18	7.79	7.70	8.68	9.75	5.98	7.41	7.65	A, B
ROM (°)	7.81	5.40	8.16	4.59	7.63	4.02	8.11	4.21	7.62	3.89
Transverse plane										
Angle at footstrike (°)	4.54	11.41	3.33	11.42	6.03	10.87	3.19	12.37	3.97	11.71
Peak external rotation (°)	-7.67	12.12	-7.43	12.63	-5.91	11.27	-9.01	12.78	-7.57	12.76
ROM (°)	12.22	6.11	10.76	6.29	11.95	6.70	12.20	6.55	11.54	5.06
	Female									
	Medial		**Lateral**		**No-orthoses**		**Semi-custom**		**Off the shelf**	
	Mean	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*
Sagittal plane										
Angle at footstrike (°)	46.66	9.69	47.82	11.08	46.53	10.75	46.03	12.02	46.24	11.90
Peak flexion (°)	47.15	9.52	48.41	10.56	47.05	10.00	47.00	11.04	47.07	10.84
ROM (°)	0.49	0.86	0.59	1.20	0.52	1.35	0.97	1.92	0.83	2.19
Coronal plane										
Angle at footstrike (°)	4.75	5.87	4.40	5.76	3.56	6.23	3.70	5.99	3.78	5.62
Peak adduction (°)	12.42	4.93	14.18	4.39	12.73	4.65	13.31	4.35	12.25	4.24	A, B
ROM (°)	7.67	3.05	9.78	4.00	9.17	3.53	9.61	3.74	8.47	4.01
Transverse plane										
Angle at footstrike (°)	10.86	8.21	10.52	8.32	10.01	7.35	10.06	8.92	11.23	8.97

Peak external rotation (°)	-2.66	7.98	-3.12	7.96	-1.68	7.72	-3.33	7.73	-2.80	8.11	
ROM (°)	13.52	6.42	13.63	6.67	11.69	6.30	13.39	6.54	14.04	7.54	

670 Key: A = main effect of ORTHOSES & B = main effect of GENDER

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693 Table 5: Three-dimensional knee joint kinematics (Mean & SD) for each orthotic condition

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	Medial		Lateral		No-orthoses		Semi-custom		Off the shelf		
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	
Sagittal plane											
Angle at footstrike (°)	14.66	5.66	16.21	6.05	13.92	6.54	15.27	6.29	14.34	6.64	B
Peak flexion (°)	43.35	6.04	44.28	6.05	42.71	5.89	43.79	5.13	43.48	5.76	B
ROM (°)	28.69	4.70	28.07	4.33	28.79	4.94	28.52	6.05	29.14	4.83	
Coronal plane											
Angle at footstrike (°)	1.22	4.96	1.06	4.15	1.58	4.96	0.44	4.55	1.60	4.82	
Peak adduction (°)	-5.96	5.37	-6.24	5.67	-5.27	4.85	-6.64	5.48	-5.49	5.33	A
ROM (°)	7.18	3.06	7.29	3.69	6.85	4.05	7.07	3.02	7.09	2.60	
Transverse plane											
Angle at footstrike (°)	-12.87	8.16	-11.55	6.23	-15.75	7.95	-11.96	8.11	-13.42	8.55	
Peak external rotation (°)	7.50	9.35	8.24	9.18	8.01	8.54	8.36	9.80	7.96	8.78	
ROM (°)	20.38	5.41	19.80	6.16	23.76	5.73	20.32	6.95	21.38	5.75	
	Female										
	Medial		Lateral		No-orthoses		Semi-custom		Off the shelf		
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	Mean	SD	
Sagittal plane											
Angle at footstrike (°)	22.57	7.86	22.47	8.14	22.85	9.89	20.63	9.46	21.37	9.58	B
Peak flexion (°)	49.92	7.93	50.79	6.93	49.19	7.06	50.26	7.66	49.75	7.52	B
ROM (°)	27.35	6.68	28.32	6.90	26.34	7.86	29.62	7.77	28.38	8.40	
Coronal plane											
Angle at footstrike (°)	0.86	5.54	1.03	6.04	1.57	5.87	0.63	5.61	1.07	5.78	
Peak adduction (°)	-6.89	4.76	-7.31	5.18	-6.19	3.65	-7.14	4.78	-6.86	4.49	A
ROM (°)	7.75	4.37	8.34	4.79	7.76	4.75	7.76	4.45	7.93	4.85	
Transverse plane											
Angle at footstrike (°)	-11.93	4.86	-12.41	7.30	-10.95	5.51	-11.46	6.97	-11.84	6.57	
Peak external rotation (°)	3.63	5.74	4.13	6.01	3.79	5.94	4.30	6.06	4.28	5.59	

ROM (°)	15.57	5.78	16.54	6.12	14.73	6.11	15.76	6.36	16.12	6.94

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738   Table 6: Three-dimensional ankle joint kinematics (Mean & SD) for each orthotic condition.

	Male										
	Medial		Lateral		No-orthoses		Semi-custom		Off the shelf		
	Mean	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*	
Sagittal plane											
Angle at footstrike (°)	2.27	15.37	2.28	16.05	-1.87	15.77	-2.21	16.39	-0.51	16.40	B
Peak dorsiflexion (°)	17.68	6.95	16.91	5.58	15.48	4.49	15.86	4.82	15.60	4.36	A
ROM (°)	15.41	11.26	14.64	12.56	17.36	14.02	18.06	15.52	16.11	13.77	B
Coronal plane											
Angle at footstrike (°)	1.93	5.00	4.40	3.98	4.68	3.91	3.64	4.04	3.75	3.91	
Peak eversion (°)	-6.61	3.69	-9.61	4.29	-8.22	3.74	-8.25	3.71	-7.54	3.74	A
ROM (°)	8.53	7.08	14.01	5.55	12.89	4.95	11.89	4.53	11.30	4.93	A
Transverse plane											
Angle at footstrike (°)	-1.78	3.12	-2.01	3.56	0.62	4.59	-0.59	3.54	0.03	3.59	
Peak external rotation (°)	-9.53	4.90	-11.04	5.05	-9.48	5.53	-10.07	5.04	-8.99	5.31	
ROM (°)	7.76	4.55	9.03	5.29	10.10	4.78	9.47	4.48	9.02	4.98	
Tibial internal rotation at footstrike (°)	8.47	5.58	8.16	5.51	7.11	5.52	8.00	5.48	7.28	5.54	
Peak tibial internal rotation (°)	15.73	5.10	17.40	5.54	15.85	5.54	16.43	5.66	15.40	5.34	A
Peak tibial internal rotation ROM (°)	7.27	3.46	9.24	4.79	8.74	3.61	8.42	3.86	8.12	4.13	A
	Female										
	Medial		Lateral		No-orthoses		Semi-custom		Off the shelf		
	Mean	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*	*Mean*	*SD*	
Sagittal plane											
Angle at footstrike (°)	11.06	6.08	11.78	5.99	10.33	5.95	10.73	5.60	11.81	5.04	B
Peak dorsiflexion (°)	20.00	3.42	18.90	3.78	19.00	3.46	18.63	3.94	18.26	4.29	A
ROM (°)	8.94	4.42	7.12	3.64	8.67	5.33	7.90	4.16	6.45	3.73	B
Coronal plane											

Angle at footstrike (°)	-0.95	5.67	3.56	6.44	2.61	5.03	2.89	5.85	1.78	5.63	
Peak eversion (°)	-7.72	4.75	-10.32	5.61	-10.10	4.04	-9.49	5.93	-8.85	4.98	A
ROM (°)	6.77	3.45	13.88	4.38	12.71	3.37	12.38	4.41	10.63	4.08	A
Transverse plane										
Angle at footstrike (°)	-4.08	6.77	-4.89	7.03	-3.30	6.89	-3.48	6.21	-3.75	6.22	
Peak external rotation (°)	-8.27	7.73	-10.44	7.21	-9.22	7.30	-10.10	7.37	-9.06	6.64	
ROM (°)	4.19	3.09	5.55	2.93	5.92	3.64	6.62	3.56	5.31	3.25	
Tibial internal rotation at footstrike (°)	11.84	6.19	11.96	6.47	10.66	6.73	10.62	6.53	10.97	5.72	
Peak tibial internal rotation (°)	16.99	6.56	19.84	6.00	18.28	6.07	19.13	6.08	18.01	5.36	A
Peak tibial internal rotation ROM (°)	5.15	3.07	7.88	2.85	7.62	3.31	8.51	3.21	7.04	3.14	A

Key: A = main effect of ORTHOSES & B = main effect of GENDER