Central Lancashire Online Knowledge (CLoK)

Title	Effects of running with minimal and conventional footwear in habitual and non-habitual users: a musculoskeletal simulation and statistical parametric mapping based approach.
Type	Article
URL	https://clok.uclan.ac.uk/30249/
DOI	https://doi.org/10.1080/19424280.2019.1683619
Date	2020
Citation	Sinclair, Jonathan Kenneth, Taylor, Paul John and Liles, Naomi (2020) Effects of running with minimal and conventional footwear in habitual and non-habitual users: a musculoskeletal simulation and statistical parametric mapping based approach. Footwear Science, 12 (1). pp. 25-38. ISSN 1942-4280
Creators	Sinclair, Jonathan Kenneth, Taylor, Paul John and Liles, Naomi

1 **Effects of running with minimal and conventional footwear in habitual and non-**

2 **habitual users; a musculoskeletal simulation and statistical parametric mapping based**

3 **approach.**

4 *Jonathan Sinclair[1], Paul John Taylor[2] & Naomi Liles[1]*

5 1. *Centre for Applied Sport and Exercise Sciences, School of Sport and Health Sciences,*

6 *Faculty of Health and Wellbeing, University of Central Lancashire.*

7 2. *School of Psychology, Faculty of Science & Technology, University of Central*

8 *Lancashire.*

9 **Keywords:** Biomechanics; musculoskeletal; footwear; running, minimal footwear.

10

11 **Abstract**

12 The current investigation examined running biomechanics in minimal and conventional

13 footwear in two groups of runners who either ran habitually in minimal footwear (habitual

14 minimal footwear users) or habitually in conventional footwear (non-habitual minimal footwear

15 users). We studied ten male non-habitual minimal footwear users and ten male habitual minimal

16 footwear users, who were required to complete ≥35 km per week of training. Lower extremity

17 joint loading was explored using a musculoskeletal simulation approach. Differences between

18 conditions were examined using statistical parametric mapping and 2x2 mixed ANOVA. This

19 study revealed via the strike index that minimal footwear caused a more anterior contact

20 position in both groups (habitual: minimal=61.68% & conventional=46.48% /non-habitual:

21 minimal=33.79% & conventional=22.61%), although non-habitual runners still adopted a

22 rearfoot strike pattern. In addition, in non-habitual users minimal footwear increased tibial

23 accelerations (habitual: minimal=6.35g & conventional=7.06g /non-habitual: minimal=9.54g

24 & conventional=8.16g), loading rates (habitual: minimal=105.44BW/s &

25 conventional=105.97BW/s /non-habitual: minimal=293.00BW/s &

26 conventional=154.36BW/s) and medial tibiofemoral loading rates (habitual:

27 minimal=196.17BW/s & conventional=274.96BW/s /non-habitual: minimal=274.96BW/s &

28 conventional=212.57BW/s). Furthermore, minimal footwear decreased patellofemoral loading

29 in both habitual (minimal=0.28BW·s & conventional=0.31BW·s) and non-habitual

30 (minimal=0.26BW·s & conventional=0.29BW·s) users. Finally, Achilles tendon loading was

31 larger in minimal footwear and in habitual runners (habitual: minimal=0.79BW·s &

32 conventional=0.71BW·s /non-habitual: minimal=0.71BW·s & conventional=0.65BW·s)

33 whereas iliotibial band strain rate was reduced in habitual (minimal=28.32%/s &

34 conventional=30.30%/s) in relation to non-habitual (minimal=42.96%/s &

35 conventional=42.87%/s) users. This study highlights firstly the importance of transitioning to

36 minimal footwear and also indicates that post transition they may be effective in attenuating the

37 biomechanical mechanisms linked to the aetiology of many chronic injuries.

38

Introduction

40 Recreational distance running is arguably the most popular aerobic exercise modality (Lee et

41 al. 2014). There is a plethora of evidence indicating that running mediates significant

42 physiological and psychological benefits (Lee et al. 2014). However, despite the physical

43 benefits that it manifests, distance running is also associated with a high susceptibility to

44 chronic injuries; as 19.4-79.3 % of runners will experience a pathology each year (Van Gent et

45 al. 2007). Unfortunately, chronic pathologies are a significant barrier to training adherence in

46 runners and lead to a substantial economic burden (Hespanhol et al. 2016). Specifically,

47 patellofemoral pain, iliotibial band syndrome, tibial stress fractures, medial tibial stress

48 syndrome, Achilles tendinopathy and pain secondary to hip and knee osteoarthritis are

49   commonly experienced in sports medicine clinics (Taunton et al. 2002, Van Ginckel et al. 2009;

50   Winkelmann et al. 2016; Snyder et al. 2006).

51

52   The running shoe is the primary interface between the body and surface; as such significant

53   developments in running shoe technology have emerged, in an attempt to mediate the incidence

54   of chronic running pathologies (Sinclair et al. 2013a). However, since the introduction of the

55   conventional running shoe, the rate and location of chronic running injuries has not changed,

56   leading to the notion that technological developments in running footwear have not been

57   successful in influencing running pathologies (Davis, 2014). This has led to the proposal that

58   running in minimal footwear that lacks the cushioning and motion control properties associated

59   with the conventional running shoe, may be associated with a reduced incidence of chronic

60   running injuries (Lieberman et al. 2010; Davis, 2014). Based on this notion, several minimal

61   footwear models are currently available commercially.

62

63   Several studies have examined differences in running biomechanics between minimal and

64   conventional running shoes. These investigations have shown that minimal footwear alter

65   spatiotemporal running characteristics, causing runners to adopt a more plantarflexed ankle at

66   footstrike (Sinclair et al. 2013ab, Hollander et al. 2015), mid/forefoot strike pattern (Squadrone

67   et al. 2015; Sinclair et al. 2019), increased stride rate (Warne et al. 2014) and reduced stride

68   length (Sinclair et al. 2016; Sinclair et al. 2019) compared to conventional running shoes. In

69   addition, previous comparisons of conventional and minimal footwear have also shown that

70   minimal footwear are associated with increased vertical loading rates (Sinclair et al. 2013ab),

71   tibial accelerations (Sinclair et al. 2013a), and effective mass (Sinclair et al. 2018a). Finally,

72   previous work examining the effects of minimal footwear on the loads experienced by the lower

73   extremities have revealed that minimal footwear reduces the loads experienced by the

74 patellofemoral joint (Sinclair, 2014; Bonacci et al. 2014), but increase the forces borne by the

75 Achilles tendon (Sinclair, 2014; Sinclair et al. 2019) and the tibiofemoral joint (Sinclair et al.

76 2018b). However, it is important to recognise that the conclusions drawn from the

77 aforementioned investigations were based upon results obtained from novice users of minimal

78 footwear. Indeed, Tam et al. (2017) proposed that in acute investigations of minimal footwear,

79 runners do not sufficiently alter their running mechanics sufficiently to reduce the vertical

80 loading rate. Therefore, it can be concluded that the overall evidence that minimal footwear is

81 able to attenuate the biomechanical factors linked to the aetiology of chronic pathologies is

82 currently insufficient. As such, with regards to minimal footwear, runners must select footwear

83 based on the findings from acute studies conducted on runners who are unaccustomed to using

84 minimal footwear. Therefore, it can be concluded that further investigation of running

85 biomechanics between minimal and conventional footwear in those who habitually wear

86 minimal and conventional footwear is warranted.

87

88 Furthermore, previous analyses concerning the biomechanical differences between minimal

89 and conventional footwear, have adopted inverse-dynamic driven modelling-based approaches

90 to quantify lower extremity musculoskeletal loading (Sinclair et al., 2019). However, joint

91 torques are representative of global indices of joint loading, and therefore are not representative

92 of localized joint loading (Herzog et al. 2003). Substantial developments in musculoskeletal

93 modelling have been made in recent years, allowing indices of skeletal muscle forces; muscle

94 kinematics and joint reaction forces be obtained through musculoskeletal simulation analyses

95 (Delp et al. 2007). This approach may be more effective than traditional inverse-dynamic based

96 methods and allows a more detailed examination of the specific parameters linked to the

97 aetiology of chronic pathologies to be undertaken. Such approaches have not yet been utilized

98 to explore biomechanical differences between minimal and conventional running shoes in

99 runners who run habitually in minimal footwear (habitual minimal footwear users) or

100 conventional footwear (non-habitual minimal footwear users).

101

102 There has yet to be a published investigation examining differences in running biomechanics

103 between minimal and conventional footwear in those who habitually wear minimal and

104 conventional footwear. Therefore, the aim of the current investigation was to examine running

105 biomechanics in minimal and conventional footwear in those who habitually wear minimal and

106 conventional footwear, with reference to the biomechanical mechanisms linked to the aetiology

107 of chronic pathologies, using a musculoskeletal simulation-based analysis.

108

109 **Methods**

110 *Participants*

111 Ten male conventional footwear users (henceforth termed non-habitual minimal footwear

112 users) (age 27.67 ± 5.57 years, height 1.71 ± 0.03 m and body mass 68.76 ± 4.78 kg) and ten

113 male habitual minimal footwear users (henceforth termed habitual minimal footwear users)

114 (age 33.50 ± 4.58 years, height 1.75 ± 0.04 m and body mass 71.74 ± 7.74 kg) volunteered to

115 take part in this study. Participants were required to complete a minimum of 35 km per week

116 of training. To be considered a habitual minimal footwear user, volunteers were required to

117 have been training exclusively in minimal footwear for a minimum period of 24 months in

118 footwear scoring ≥ 75 on the minimalist index described by Esculier et al. (2015). The

119 procedure utilized for this investigation was approved by a university ethical committee (REF

120 637). All runners were free from musculoskeletal pathology at the time of data collection and

121 provided written informed consent in accordance with the principles outlined in the Declaration

122 of Helsinki.

123

124 *Footwear*

125 The footwear used during this study consisted of New Balance, 1260 v2 (New Balance, Boston,

126 Massachusetts, United States; henceforth termed conventional) and Vibram Five-Fingers, ELX

127 (Vibram, Albizzate, Italy; henceforth termed minimal) (Figure 1). The conventional footwear

128 had an average mass of 0.285 kg, heel thickness of 25 mm and a heel drop of 14 mm and

129 minimal an average mass of 0.167 kg, heel thickness of 7 mm and a heel drop of 0 mm. The

130 footwear were also scored using the minimalist index of Esculier et al. (2015), and the

131 conventional footwear received a score of 20 and minimal a score of 92.

132

133   **@@@FIGURE 1 NEAR HERE@@@**

134

135 *Procedure*

136 Participants ran at 4.0 m/s (± 5%), striking an embedded piezoelectric force platform (Kistler

137 Instruments Ltd., Winterthur, Switzerland) with their right (dominant) foot. Running velocity

138 was monitored using infrared timing gates (Newtest, Oy Koulukatu, Finland). The stance phase

139 was delineated as the duration over which 20 N or greater of vertical ground reaction force

140 (GRF) was applied to the force platform. Runners completed five successful trials in each

141 footwear condition. The order that participants ran in each footwear condition was

142 counterbalanced. Kinematic and GRF data were synchronously collected. Kinematic data were

143 captured at 250 Hz via an eight-camera motion analysis system (Qualisys Medical AB,

144 Goteburg, Sweden). Dynamic calibration of the motion capture system was performed before

145 each data collection session.

146

147 Body segments were modelled in 6 degrees of freedom using the calibrated anatomical systems

148 technique (Cappozzo et al. 1995). To define the anatomical frames of the thorax, pelvis, thighs,

149 shanks and feet retroreflective markers were placed at the C7, T12 and xiphoid process

150 landmarks and also positioned bilaterally onto the acromion process, iliac crest, anterior

151 superior iliac spine (ASIS), posterior super iliac spine (PSIS), medial and lateral malleoli,

152 medial and lateral femoral epicondyles, greater trochanter, calcaneus, first metatarsal and fifth

153 metatarsal. Carbon-fibre tracking clusters comprising of four non-linear retroreflective markers

154 were positioned onto the thigh and shank segments. In addition to these, the foot segments were

155 tracked via the calcaneus, first metatarsal and fifth metatarsal, the pelvic segment was tracked

156 using the PSIS and ASIS markers and the thorax segment was tracked using the T12, C7 and

157 xiphoid markers. Static calibration trials were obtained in each footwear allowing for the

158 anatomical markers to be referenced in relation to the tracking markers/ clusters.

159

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

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

162 of lightweight carbon-fibre material using the protocol outlined by Sinclair et al. (2013a). The

163 accelerometer was attached securely to the distal anterio-medial aspect of the tibia in alignment

164 with its longitudinal axis, 0.08 m above the medial malleolus. Strong non-stretch adhesive tape

165 was placed over the device and leg to avoid overestimating the acceleration due to tissue

166 artefact (Sinclair et al. 2013a).

167

168 *Processing*

169 Dynamic trials were digitized using Qualisys Track Manager (Qualisys Medical AB, Goteburg,

170 Sweden) in order to identify anatomical and tracking markers then exported as C3D files to

171 Visual 3D (C-Motion, Germantown, MD, USA). All data were linearly normalized to 100 %

172   of the stance phase. GRF data and marker trajectories were smoothed with cut-off frequencies

173   of 50 Hz at 12 Hz respectively, using a low-pass Butterworth 4th order zero lag filter. In

174   addition, the tibial acceleration signal was filtered using a 60 Hz Butterworth zero lag 4th order

175   low pass filter (Sinclair et al. 2013a). Kinematics of the hip, knee and ankle were quantified

176   using an XYZ cardan sequence of rotations (where X is flexion-extension; Y is ab-adduction

177   and is Z is internal-external rotation). In addition, tibial internal rotation kinematics were also

178   calculated in accordance with Eslami et al. (2007). All force parameters throughout were

179   normalized by dividing by bodyweight (BW).

180

181   In accordance with the protocol of Addison & Lieberman, (2015), an impulse-momentum

182   modelling approach was utilized to calculate effective mass (% BW), which was quantified in

183   accordance with the below equation:

184

185   Effective mass = vertical GRF integral / (*Δ foot vertical velocity + gravity * Δ time)*

186

187   The impact peak was defined firstly in non-habitual runners when wearing conventional

188   footwear, as the first peak in vertical GRF. In habitual runners and non-habitual runners

189   wearing minimal footwear where no impact peak was expected, according to the protocols of

190   Lieberman et al. (2010) and Sinclair et al. (2018a) we defined the position of the impact peak

191   at the same relative position, which was shown to be 11.87 % of the stance phase. The time

192   (ms) to impact peak (*Δ time*) was quantified as the duration from footstrike to impact peak. The

193   vertical GRF integral (BW·ms) during the period of the impact peak was calculated using a

194   trapezoidal function. The change in foot vertical velocity (*Δ foot vertical velocity*) was

195   determined as the instantaneous vertical foot velocity averaged across the 10 frames prior to

196   the impact peak (Sinclair et al. 2018a). The velocity of the foot was quantified using the centre

197   of mass of the foot segment in the vertical direction, within Visual 3D (Sinclair et al. 2018a).

198

199   Loading rate (BW/s) was also was also extracted by obtaining the peak increase in vertical

200   GRF between adjacent data points using the first derivative function within Visual 3D and the

201   peak tibial acceleration (g) was extracted as the highest positive acceleration peak during the

202   stance phase. The strike index was calculated as the position of the centre of pressure location

203   at footstrike, relative to the total length of the foot (Squadrone et al. 2015). A strike index of

204   0–33% denotes a rearfoot, 34–67% a midfoot and 68–100% a forefoot strike pattern. Finally,

205   limb stiffness during running was quantified using a mathematical spring-mass model

206   (Blickhan, 1989). Limb stiffness (BW/m) was calculated from the ratio of the peak normalized

207   vertical GRF to the maximum vertical compression of the leg spring which was calculated as

208   the change in limb length from footstrike to minimum length during the stance phase (Farley

209   & Morgenroth, 1999). Limb length was quantified as the vertical height of the proximal end of

210   the thigh segment within Visual 3D.

211

212   Following this, data during the stance phase were exported from Visual 3D into OpenSim 3.3

213   software (Simtk.org). Two validated musculoskeletal models were used to process the

214   biomechanical data both of which were scaled to account for the anthropometrics of each

215   runner. The first with 12 segments, 19 degrees of freedom and 92 musculotendon actuators

216   (Lerner et al. 2015) was used initially to estimate lower extremity joint forces. As muscle forces

217   are the main determinant of joint compressive forces (Herzog et al. 2003), muscle kinetics were

218   quantified using static optimization in accordance with Steele et al. (2012). Compressive

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

220 joint reaction analyses function using the muscle forces generated from the static optimization

221 process as inputs. Furthermore, patellofemoral stress (KPa/kg) was quantified by dividing the

222 patellofemoral force by the contact area. Patellofemoral contact areas were obtained by fitting

223 a polynomial curve to the sex specific data of Besier et al. (2005), who estimated patellofemoral

224 contact areas as a function of the knee flexion angle using MRI. Finally, Achilles tendon forces

225 were estimated in accordance with the protocol of Almonroeder et al. (2013), by summing the

226 muscle forces of the medial gastrocnemius, lateral, gastrocnemius, and soleus muscles.

227

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

229 instantaneous load rates (BW/s and KPa/BW/s) were also extracted by obtaining the maximum

230 increase in force/ stress between adjacent data points using the first derivative function in Visual

231 3D. Finally, the integral of the hip, tibiofemoral, ankle, patellofemoral and Achilles tendon

232 forces (BW·s) and stresses (KPa/BW·s) during the stance phase were calculated using a

233 trapezoidal function.

234

235 Running in minimal footwear has been shown to alter step length during running, which

236 increases the number of footstrikes necessary to run a set distance. We therefore estimated the

237 total impulse per kilometre (BW·km) by multiplying these parameters by the number of steps

238 required to run a kilometre. The number of steps required to complete one kilometre was

239 quantified using the step length (m), which was determined by taking the difference in the

240 horizontal position of the foot centre of mass between the right and left legs at footstrike.

241

242 The second model also had twelve segments, 23 degrees of freedom and 92 muscle-tendon

243 actuators and was adapted from the generic OpenSim gait2392 model to include the iliotibial

244 band (Foch et al. 2013). The iliotibial band itself was included within the gait2392 model but

245   as a muscle with only a passive contractile component and an optimal muscle fiber length of

246   zero (Foch et al. 2013). Iliotibial band kinematics during the stance phase were calculated via

247   the muscle analyses function within OpenSim and iliotibial band strain (%) was calculated by

248   dividing the change in length of the band during stance and dividing by its resting length at each

249   time frame. In addition, the strain rate (%/s) was calculated as the change in strain between

250   adjacent data points. The resting length of the iliotibial band was determined as its length during

251   the static calibration trial (Hamill et al. 2008). Peak iliotibial band strain and strain rate were

252   measured at the instance of peak knee flexion during stance (Hamill et al. 2008).

253

254   *Statistical analyses*

255   Following data processing, compressive joint forces (hip, patellofemoral, medial tibiofemoral

256   and lateral tibiofemoral), Achilles tendon loading and three-dimensional kinematics during the

257   entire stance phase were temporally normalized using linear interpolation to 101 data points.

258   Differences across the entire stance phase were examined using 1-dimensional statistical

259   parametric mapping (SPM) with MATLAB 2017a (MATLAB, MathWorks, Natick, USA), in

260   accordance with Pataky et al. (2016), using the source code available at

261   http://www.spm1d.org/. Differences as a function of both FOOTWEAR (FOOTWEAR –

262   conventional or minimal) and GROUP (GROUP - habitual or non-habitual) were examined

263   using paired and independent t-tests (SPM t).

264

265   For discrete parameters that could not be examined using SPM (joint integral, joint loading

266   rate, joint integral per kilometre, step length, instantaneous load rate, strike index, limb

267   stiffness, tibial accelerations, iliotibial band strain, iliotibial band strain rate and effective

268   mass), means and standard deviations were calculated for each condition. Differences in

269   discrete biomechanical parameters were examined using 2 (FOOTWEAR – conventional of

270 minimal) x 2 (GROUP- habitual or non-habitual) mixed ANOVAs, Effect sizes were calculated

271 using partial eta^2 (pη^2). In the event of a significant interaction, simple main effects tests were

272 adopted. Discrete statistical actions were conducted using SPSS v25.0 (SPSS Inc., Chicago,

273 USA). Statistical significance was accepted at the P≤0.05 level.

274

275 **Results**

276 @@@**TABLE 1 NEAR HERE**@@@

277 @@@**TABLE 2 NEAR HERE**@@@

278

279 *Lower extremity external loading, strike index and step length*

280 For effective mass there was a significant FOOTWEAR*GROUP interaction (P=0.01, pη^2 =

281 0.31). Simple main effects tests showed that effective mass was larger in the conventional

282 running shoes compared to minimal in habitual runners (P=0.01, pη^2 = 0.53) but there were no

283 significant differences between footwear in non-habitual runners (P=0.26, pη^2 = 0.11). In

284 addition, when wearing minimal footwear, effective mass was significantly greater in non-

285 habitual runners compared to habitual (P<0.001, pη^2 = 0.61) but there were no differences

286 between habitual and non-habitual runners when running in conventional footwear (P=0.50,

287 pη^2 = 0.03) (Table 1).

288

289 For loading rate there was also a significant FOOTWEAR*GROUP interaction (P=0.002, pη^2

290 = 0.41). Simple main effects tests showed that loading rate was significantly larger in the

291 minimal footwear compared to conventional in non-habitual runners (P=0.004, pη^2 = 0.63) but

292 there was no significant difference between footwear in habitual runners (P=0.94, pη^2 <0.001).

293 In addition, when wearing minimal footwear, the loading rate was significantly greater in non-

294 habitual runners compared to habitual (P<0.001, $p\eta^2 = 0.52$) but there were no differences when

295 running in conventional footwear (P=0.06, $p\eta^2 = 0.19$) (Table 1).

296

297 For peak tibial accelerations, there was a significant FOOTWEAR*GROUP interaction

298 (P=0.005, $p\eta^2 = 0.36$). Simple main effects tests showed that tibial accelerations were

299 significantly larger in minimal footwear compared to conventional in non-habitual runners

300 (P=0.03, $p\eta^2 = 0.42$) but there was no significant difference between fotowear in habitual

301 runners (P=0.09, $p\eta^2 = 0.29$). In addition, when wearing minimal footwear, tibial accelerations

302 were significantly greater in non-habitual compared to habitual runners (P<0.001, $p\eta^2 = 0.57$)

303 but there were no differences between habitual and non-habitual runners in conventional

304 footwear (P=0.20, $p\eta^2 = 0.09$) (Table 1).

305

306 For limb stiffness there was a significant FOOTWEAR*GROUP interaction (P=0.04, $p\eta^2 =$

307 0.21). Simple main effects tests showed that limb stiffness was greater in minimal compared to

308 conventional footwear in non-habitual runners (P<0.001, $p\eta^2 = 0.57$) but there were no

309 differences between footwear when running in conventional footwear (P=0.20, $p\eta^2 = 0.09$)

310 (Table 1).

311

312 For strike index there was a main effect of FOOTWEAR (P=0.002, $p\eta^2 = 0.36$), which showed

313 that the strike position was more anterior in minimal footwear. In addition, there was also a

314 main effect of GROUP (P=0.007, $p\eta^2 = 0.34$), which indicated that the strike was also more

315 anterior in habitual runners (Table 1).

316

317 For step length there was a significant FOOTWEAR*GROUP interaction (P=0.04, $p\eta^2 = 0.20$).

318 Simple main effects tests showed that step length was significantly larger in conventional

319 compared to minimal footwear in habitual runners (P=0.001, $p\eta^2 = 0.72$) but there was no

320 difference between footwear in non-habitual runners (P=0.70, $p\eta^2 = 0.02$). In addition, when

321 wearing minimal footwear compared to conventional, step length was significantly greater in

322 non-habitual runners (P=0.02, $p\eta^2 = 0.28$) but there were no differences between habitual and

323 non-habitual runners when running in conventional footwear (P=0.11, $p\eta^2 = 0.14$) (Table 1).

324

325 *Joint loading*

326 For medial tibiofemoral loading rate there was a significant FOOTWEAR*GROUP interaction

327 (P<0.001, $p\eta^2 = 0.76$). Simple main effects tests showed that the loading rate was significantly

328 larger in the conventional compared to minimal footwear in habitual runners (P=0.001, $p\eta^2 =$

329 0.91) but significantly greater in minimal compared to conventional footwear in non-habitual

330 runners (P=0.005, $p\eta^2 = 0.61$). In addition, when wearing minimal footwear, medial

331 tibiofemoral loading rate was significantly greater in non-habitual compared to habitual runners

332 (P=0.02, $p\eta^2 = 0.26$) but in conventional footwear was significantly greater in habitual

333 compared to non-habitual runners (P=0.04, $p\eta^2 = 0.21$) (Table 1).

334

335 For the integral of patellofemoral joint force, there was a main effect of FOOTWEAR (P=0.03,

336 $p\eta^2 = 0.25$), which was shown to be larger in conventional footwear (Table 1).

337

338 For the integral of Achilles tendon force, there was a main effect of FOOTWEAR (P=0.02, $p\eta^2$

339 $= 0.27$), which was shown to be larger in minimal footwear. In addition, there was a main effect

340 for GROUP (P=0.002, $p\eta^2 = 0.42$), which indicated that the Achilles tendon integral was greater

341 in habitual runners (Table 1). For the Achilles tendon integral per kilometre, there was a main

342   effect of FOOTWEAR (P=0.004, pη^2 = 0.38), which was shown to be larger in minimal

343   footwear. In addition, there was a main effect for GROUP (P=0.002, pη^2 = 0.41), which

344   indicated that the Achilles tendon integral was greater in habitual runners (Table 2).

345

346   For the ankle integral per kilometre, there was a main effect for GROUP (P=0.02, pη^2 = 0.27),

347   which indicated that the ankle integral was greater in habitual runners (Table 2).

348

349   *Iliotibial band kinematics*

350   For iliotibial band strain rate, there was a main effect for GROUP (P<0.001, pη^2 = 0.52), which

351   indicated that the strain rate was greater in non-habitual runners (Table 1).

352

353   *Statistical parametric mapping - joint loading*

354   Minimal footwear was associated with increased Achilles tendon force compared to

355   conventional running shoes in the first 20% of the stance phase in both habitual and non-

356   habitual runners (Figure 2ab).

357

358   *Statistical parametric mapping - three-dimensional kinematics*

359   Conventional footwear was associated with increased hip flexion compared to minimal from

360   20-40% of the stance phase in habitual runners (Figure 2c). Conventional footwear was also

361   associated with increased knee flexion compared to minimal from 40-60% of the stance phase

362   in both habitual and non-habitual runners (Figure 2de). In additional, minimal footwear

363   compared to conventional was associated with increased tibial and knee internal rotation during

364   from 20-60% of the stance phase in habitual runners (Figure 3ab). Furthermore, it was revealed

365   that the ankle exhibited increased plantarflexion in minimal footwear from 0-5% of the stance

366   phase in both habitual and non-habitual runners (Figure 3cd). Finally, in conventional footwear

367   compared to minimal, habitual runners were similarly associated with increased plantarflexion

368   from 0-5% of the stance phase (Figure 3e).

369

370   **@@@FIGURE 2 NEAR HERE@@@**

371   **@@@FIGURE 3 NEAR HERE@@@**

372

373   **Discussion**

374   The aim of the current investigation was to examine differences in running biomechanics

375   between minimal and conventional footwear, in those who habitually wear minimal and

376   conventional footwear. To the authors knowledge, this is the first quantitative comparison of

377   these footwear in habitual and non-habitual minimal footwear users using a musculoskeletal

378   simulation and SPM based approach.

379

380   The kinematic analysis using SPM of the sagittal plane ankle angle aligned with the discrete

381   analysis of the strike index, supports previous investigations in that minimal footwear

382   transferred the footstrike to a more anterior position in both habitual and non-habitual runners

383   (Squadrone et al. 2015; Sinclair et al. 2019). Furthermore, in support of previous analyses the

384   findings from this study also showed that habitual minimal footwear users similarly were

385   associated with a significantly more anterior footsrike position in relation to non-habitual

386   runners (Larson et al. 2014). It is important to contextualize the strike index values observed in

387   both conditions, as regardless of which footwear condition was utilized non-habitual runners

388   maintained a rearfoot strike pattern and habitual runners adopted a midfoot contact position.

389   This supports proposition of Tam et al. (2017) that in acute investigations non-habitual runners

390   do not sufficiently alter their running mechanics and continue to exhibit a rearfoot strike pattern.

391

392   For the indices of external loading, in agreement with previous analyses this investigation

393   showed that tibial accelerations and loading rates were found to be greater in minimal footwear

394   in non-habitual runners (Sinclair et al. 2013ab) and in non-habitual runners when wearing

395   minimal footwear (Lieberman et al. 2010). As non-habitual runners adopted a rearfoot strike

396   pattern when wearing minimal footwear, it was expected that both effective mass and limb

397   stiffness were also increased when non-habitual runners adopted minimal footwear. It is

398   proposed that the increases in external loading indices were mediated by the corresponding

399   changes in effective mass and limb stiffness, which have been shown previously to be positively

400   related to the magnitude of the both tibial accelerations and loading rate (Sinclair et al. 2018a).

401   As tibial accelerations/ loading rates were increased in non-habitual runners using minimal

402   footwear, these observations may be clinically meaningful. Given the proposed association

403   between tibial accelerations/ loading rates and the aetiology of chronic injuries (Davis et al.

404   2004), this study indicates that non-habitual runners wearing minimal footwear are at increased

405   risk from impact related injuries.

406

407   Although no differences were revealed using SPM, the discrete analysis showed that the

408   patellofemoral force integral was significantly larger in conventional footwear in both habitual

409   and non-habitual groups. This finding concurs with those observed previously by Sinclair,

410   (2014), Sinclair et al. (2016) and Bonacci et al. (2014) who showed significant reductions in

411   patellofemoral loading when running in minimal footwear. The discrete and SPM based

412   analyses showed that minimal footwear transferred the footstrike to a more anterior position

413   and also reduced the extent of peak knee flexion in both habitual and non-habitual groups. It

414   is proposed that these observations are responsible for the reductions in patellofemoral loading

415   as previous analyses have shown that the function of the knee joint as an energy absorber is

416   reduced when there is an increased plantarflexion involvement (Sinclair & Selfe, 2015).

417   Importantly, excessive patellofemoral joint loading is considered a key mechanism linked to

418   the aetiology of pain symptoms in active individuals (Ho et al. 2012). Therefore, the findings

419   from the current investigation indicate that in both habitual and non-habitual runners, minimal

420   footwear may be effective in attenuating the biomechanical parameters linked to the aetiology

421   of patellofemoral pain.

422

423   In addition, it was revealed via the discrete analysis, that the loading rate at the medial aspect

424   of the tibiofemoral joint was larger in the conventional footwear in habitual runners and in

425   minimal footwear in non-habitual runners. This supports those of Sinclair et al. (2018b) who

426   showed in non-habitual runners, that minimal footwear increased the loading rate at the medial

427   aspect of the knee joint. This observation indicates that the loading rate at the medial

428   tibiofemoral joint was statistically larger when runners performed in their non-preferred

429   footwear condition. Because the loading rate at the medial knee has been cited as important

430   predictor of radiographic knee osteoarthritis, the findings from this investigation indicate that

431   runners are at increased risk when running in their non-preferred footwear condition without

432   habituation (Morgenroth et al. 2014).

433

434   Furthermore, this investigation showed using both SPM and discrete analyses that Achilles

435   tendon loading indices were significantly larger in minimal footwear and in habitual runners

436   collectively. This observation concurs with previous investigations (Sinclair, 2014, Sinclair et

437   al. 2019) showing that in non-habitual runners' minimal footwear significantly enhanced

438   Achilles tendon loading compared to conventional running shoes, although there is no

439   comparative literature examining the mechanics of the Achilles tendon in habitual minimal

440   footwear users. Importantly, the current study also showed that habitual runners were associated

441   with enhanced Achilles tendon loading compared to non-habitual users. It is proposed that the

442   mechanism responsible for these observations is the more anterior footsrike position in minimal

443   footwear and in habitual users, which served to enhance triceps surae muscle forces during the

444   eccentric aspect of the stance phase (Almonroeder et al. 2013). This observation may be

445   clinically important, as the initiation of Achilles tendinopathy is believed to be mediated

446   through repeated and excessive loads experienced by tendon itself without sufficient rest in

447   between loading exposures (Selvanetti et al. 1997). However, Davis et al. (2017) postulate that

448   greater tendon loading in habituated runners may instigate the stimulus required for tendon

449   hypertrophy and enhanced stiffness within the muscle–tendon unit necessary for the storage

450   and release of elastic energy. Anrsten et al. (2017) support this notion as they showed that

451   habitual minimal footwear users were associated with greater tendon cross sectional area and

452   increased stiffness.

453

454   Finally, the current study also importantly showed that iliotibial band strain rate was greater in

455   non-habitual runners. This finding may be clinically important as modelling investigations

456   suggest that increased strain rate is the biomechanical risk factor linked to the aetiology of

457   iliotibial band syndrome (Hamill et al. 2008). The main mechanical difference (irrespective of

458   footwear) between groups, was the adoption of a midfoot strike pattern in habitual minimal

459   footwear users compared to non-habitual. Therefore, the findings from this study lend support

460   to the proposition of Lalonde (2013) that a rearfoot landing should be avoided for the

461   prevention of iliotibial band syndrome in runners, although further aetiological investigations

462   are required to substantiate this notion. As such, the current investigation indicates that

463   transitioning to minimal footwear may be beneficial for runners in that they are able to attenuate

464   their risk from iliotibial band syndrome.

465

466   A potential limitation to the current study that should be acknowledged is that only male runners

467   were examined. Females have been shown to exhibit distinct external loading kinetics (Ferber

468   et al. 2003), lower extremity kinematics (Sinclair et al. 2012, Ferber et al. 2003), limb stiffness

469   (Sinclair et al. 2016), patellofemoral (Sinclair & Selfe, 2015) and Achilles tendon (Greenhalgh

470   & Sinclair, 2014), parameters compared to male runners. This therefore suggests that further

471   investigation of minimal footwear in habitual users using a female sample is warranted before

472   comprehensive conclusions can be drawn. Furthermore, the efficacy of musculoskeletal

473   simulation analyses depends on the fidelity of the primary neuromusculoskeletal model used to

474   quantify the mechanics of the movement being investigated (Seth et al., 2011). Many

475   assumptions and simplifications are made in the development of musculoskeletal simulation

476   models, which could potentially impact the results from the current investigation (Seth et al.,

477   2011). Therefore, there is considerable scope for future analyses to address and improve upon

478   these limitations, in order to provide more accurate and valid musculoskeletal simulations.

479

480   In conclusion, the biomechanics of minimal and conventional footwear have received

481   widespread research attention. However, there has not been quantitative comparison of these

482   footwear in habitual and non-habitual minimal footwear users using a musculoskeletal

483   simulation and SPM based approach. This study revealed that minimal footwear mediated a

484   more anterior contact position in both groups, although non-habitual runners still adopted a

485   rearfoot strike pattern. In addition, minimal footwear increased tibial accelerations, loading

486   rates and medial tibiofemoral loading rates in non-habitual runners and decreased

487   patellofemoral loading in both habitual and non-habitual groups. Finally, Achilles tendon

488 loading indices were larger in minimal footwear and in habitual runners whereas iliotibial band

489 strain rate was reduced in habitual runners. Therefore, this study highlights firstly the

490 importance of transitioning to minimal footwear and also indicates that post transition they may

491 be effective in attenuating the biomechanical mechanisms linked to the aetiology of many

492 chronic injuries.

493

494 **References**

495 1. Addison BJ, Lieberman DE. Tradeoffs between impact loading rate, vertical impulse

496   and effective mass for walkers and heel strike runners wearing footwear of varying

497   stiffness. J Biomech. 2015; 48: 1318-1324.

498 2. Almonroeder T, Willson JD, Kernozek TW. The effect of foot strike pattern on Achilles

499   tendon load during running. Annal Biomed Eng. 2013; 41: 1758-1766.

500 3. Besier TF, Draper CE, Gold GE, Beaupré GS, Delp SL. Patellofemoral joint contact

501   area increases with knee flexion and weight- bearing. J Orthop Res; 23; 345-350.

502 4. Blickhan R. The spring-mass model for running and hopping. J Biomech. 1989; 22:

503   1217-1227.

504 5. Bonacci J, Vicenzino B, Spratford W, Collins P. Take your shoes off to reduce

505   patellofemoral joint stress during running. Br J Sports Med. 2014; 48: 425-428.

506 6. Cappozzo A, Catani F, Leardini A, Benedeti MG, Della CU. Position and orientation

507   in space of bones during movement: Anatomical frame definition and determination.

508   Clin Biomech. 1995; 10: 171-178.

509 7. Davis IS. The re-emergence of the minimal running shoe. J Orthop Sports Phys Ther.

510   2014; 44: 775-784.

511 8. Davis I, Milner CE, Hamill, J. Does Increased Loading during Running Lead to Tibial

512   Stress Fractures? A Prospective Study. Med. Sci. Sports Exerc 2004, 36, S58,

513   9.   Delp SL, Anderson FC, Arnold AS, Loan P, Habib A, John CT, Thelen DG. OpenSim:

514   open-source software to create and analyze dynamic simulations of movement. IEEE

515   Trans Biomed Eng. 2007; 54: 1940-1950.

516   10. Esculier JF, Dubois B, Dionne CE, Leblond J, Roy JS. A consensus definition and

517   rating scale for minimalist shoes. J Foot Ankle Res. 2015; 8: 42-46.

518   11. Eslami M, Begon M, Farahpour N, Allard P. Forefoot–rearfoot coupling patterns and

519   tibial internal rotation during stance phase of barefoot versus shod running. Clin

520   Biomech. 2007; 22: 74-80.

521   12. Farley CT, Morgenroth DC. Leg stiffness primarily depends on ankle stiffness during

522   human hopping. J Biomech. 1999; 32: 267-273.

523   13. Ferber R, Davis IM, Williams DS. Gender differences in lower extremity mechanics

524   during running. Clin Biomech. 2003; 18: 350-357.

525   14. Foch E, Reinbolt JA, Zhang S, Milner CE. Association between iliotibial band

526   syndrome status and running biomechanics in women. Proceedings of American

527   Society of Biomechanics. 2013; Omaha, NE.

528   15. Greenhalgh A, Sinclair J. Comparison of Achilles tendon loading between male and

529   female recreational runners. J Human Kinet. 2014; 44: 155-159.

530   16. Herzog W, Longino D, Clark A. The role of muscles in joint adaptation and

531   degeneration. Langenbeck Arch Surg. 2003; 388: 305-315.

532   17. Hespanhol LC, Van Mechelen W, Postuma E, Verhagen E. Health and economic

533   burden of running-related injuries in runners training for an event: A prospective cohort

534   study. Scand J Med Sci Sports. 2016; 26: 1091-1099.

535   18. Hamill J, Miller R, Noehren B, Davis I. A prospective study of iliotibial band strain in

536   runners. Clin Biomech. 2008; 23: 1018-1025.

537   19. Histen K, Arntsen J, L'Hereux L, Heeren J, Wicki B, Saint S, Joseph MF. Achilles

538   tendon properties of minimalist and traditionally shod runners. J Sport Rehab. 2017;

539   26: 159-164.

540   20. Ho KY, Blanchette MG, Powers CM. The influence of heel height on patellofemoral

541   joint kinetics during walking. Gait Posture. 2012; 36: 271-275.

542   21. Hollander K, Argubi-Wollesen A, Reer R, Zech A. Comparison of minimalist footwear

543   strategies for simulating barefoot running: a randomized crossover study. PloS one.

544   2015; 10: e0125880.

545   22. Lalonde F. Iliotibial Band Syndrome: Noninvasive Solutions for Runners. ACSM's

546   Health Fitness Journal. 2013; 17: 9-13.

547   23. Lee DC, Pate RR, Lavie CJ, Sui X, Church TS, Blair SN. Leisure-time running reduces

548   all-cause and cardiovascular mortality risk. J Am Coll Cardiol. 2014; 64: 472-481.

549   24. Lerner ZF, DeMers MS, Delp SL, Browning RC. How tibiofemoral alignment and

550   contact locations affect predictions of medial and lateral tibiofemoral contact forces. J

551   Biomech. 2015; 48: 644-650.

552   25. Lieberman DE, Venkadesan M, Werbel WA, Daoud AI, D'andrea S, Davis IS,

553   Pitsiladis Y. Foot strike patterns and collision forces in habitually barefoot versus shod

554   runners. Nature. 2010; 463: 531-535.

555   26. Larson P. Comparison of foot strike patterns of barefoot and minimally shod runners in

556   a recreational road race. J Sport Health Sci. 2014; 3: 137-142.

557   27. Morgenroth DC, Medverd JR, Seyedali M, Czerniecki JM. The relationship between

558   knee joint loading rate during walking and degenerative changes on magnetic resonance

559   imaging. Clin Biomech. 2014; 29: 664-670.

560   28. Pataky TC, Robinson MA, Vanrenterghem J. Region-of-interest analyses of one-

561   dimensional biomechanical trajectories: bridging 0D and 1D theory, augmenting

562   statistical power. Peer J. 2016; 4: 2652-2664.

563   29. Selvanetti A, Cipolla M, Puddu G. Overuse tendon injuries: basic science and

564   classification. Oper Tech Sports Med. 1997; 5: 110-117.

565   30. Seth A, Sherman M, Reinbolt JA, Delp SL. OpenSim: a musculoskeletal modeling and

566   simulation framework for in silico investigations and exchange. Procedia IUTAM.

567   2011; 2: 212-232.

568   31. Sinclair J, Greenhalgh A, Edmundson CJ, Brooks D, Hobbs SJ. Gender differences in

569   the kinetics and kinematics of distance running: implications for footwear design. Int J

570   Sport Sci Eng. 2012; 6: 118-128.

571   32. Sinclair J, Greenhalgh A, Edmundson CJ, Brooks D, Hobbs SJ. The influence of

572   barefoot and barefoot-inspired footwear on the kinetics and kinematics of running in

573   comparison to conventional running shoes. Footwear Sci. 2013a; 5: 45–53.

574   33. Sinclair J, Hobbs SJ, Currigan G, Taylor PJ. A comparison of several barefoot inspired

575   footwear models in relation to barefoot and conventional running footwear. Comp

576   Exerc Physiol. 2013b; 9: 13-21.

577   34. Sinclair J. Effects of barefoot and barefoot inspired footwear on knee and ankle loading

578   during running. Clin Biomech. 2014; 29: 395-399.

579   35. Sinclair J, Selfe J. Sex differences in knee loading in recreational runners. J Biomech.

580   2015; 48: 2171-2175.

581   36. Sinclair J, Richards J, Selfe J, Fau-Goodwin J, Shore H. The influence of minimalist

582   and maximalist footwear on patellofemoral kinetics during running. J Appl Biomech.

583   2016; 32: 359-364.

584   37. Sinclair J, Shore HF, Taylor PJ, Atkins S. Sex differences in limb and joint stiffness in

585   recreational runners. Hum Mov. 2016; 16: 137-141.

586   38. Sinclair J, Stainton P, Hobbs SJ. Effects of barefoot and minimally shod footwear on

587   effective mass–Implications for transient musculoskeletal loading. Kinesiology. 2018a;

588   50: 165-171.

589   39. Sinclair J, Butters B, Stainton P. Acute effects of barefoot and minimalist footwear on

590   medial tibiofemoral compartment loading during running: A statistical parametric

591   mapping approach. J Hum Kinet. 2018b; 65: 35-40.

592   40. Sinclair J, Brooks D, Stainton P. Biomechanical effects of a lightweight, sock-style

593   minimalist footwear design during running: a musculoskeletal simulation and statistical

594   parametric mapping approach. Footwear Sci. 2019; 11: 71-83.

595   41. Snyder RA, Koester MC, Dunn WR. Epidemiology of stress fractures. Clin Sports Med.

596   2006; 25: 37-52.

597   42. Squadrone R, Rodano R, Hamill J, Preatoni E. Acute effect of different minimalist

598   shoes on foot strike pattern and kinematics in rearfoot strikers during running. J Sports

599   Sci. 2015; 33: 1196-1204.

600   43. Steele KM, DeMers MS, Schwartz MH, Delp SL. Compressive tibiofemoral force

601   during crouch gait. Gait Posture. 2012; 35: 556-560.

602   44. Tam N, Darragh IA, Divekar NV, Lamberts RP. Habitual minimalist shod running

603   biomechanics and the acute response to running barefoot. Int J Sports Med. 2017; 38:

604   770-775.

605   45. Taunton JE, Ryan MB, Clement DB, McKenzie DC, Lloyd-Smith DR, Zumbo BD. A

606   retrospective case-control analysis of 2002 running injuries. Br J Sports Med. 2002; 36:

607   95-101.

608   46. van Gent BR, Siem DD, van Middelkoop M, van Os TA, Bierma-Zeinstra SS, Koes

609   BB. Incidence and determinants of lower extremity running injuries in long distance

610   runners: a systematic review. Br J Sports Med. 2007; 41: 469-480.

611   47. Van Ginckel A, Thijs Y, Hesar NGZ, Mahieu N, De Clercq D, Roosen P, Witvrouw E.

612   Intrinsic gait-related risk factors for Achilles tendinopathy in novice runners: a

613   prospective study. Gait Posture. 2009; 29: 387-391.

614   48. Warne JP, Kilduff SM, Gregan BC, Nevill AM, Moran KA, Warrington GD. A 4-

615   week instructed minimalist running transition and gait- retraining changes plantar

616   pressure and force. Scand J Med Sci Sports. 2014; 24: 964-973.

617   49. Winkelmann ZK, Anderson D, Games KE, Eberman LE. Risk factors for medial tibial

618   stress syndrome in active individuals: an evidence-based review. J Ath Train. 2016; 51:

619   1049-1052.

620

621

622

623

624

625

626

627

628

629

630

631

632

Table 1: Discrete biomechanical parameters (mean ± standard deviations) as a function of FOOTWEAR
and GROUP.

	Non-habitual				Habitual				
	Conventional		Minimal		Conventional		Minimal		
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	
Effective mass (% BW)	9.59	1.93	11.32	1.81	9.06	1.53	7.83	1.01	B, C
Loading rate (BW/s)	154.36	69.86	293.00	126.14	105.97	27.20	105.44	48.95	A, B, C
Peak tibial acceleration (g)	8.16	2.04	9.54	1.90	7.06	1.60	6.35	0.86	B, C
Limb stiffness (BW/m)	63.41	28.52	65.91	22.69	63.46	33.51	48.12	13.97	C
Iliotibial band strain (%)	2.41	2.09	2.44	1.85	2.09	2.18	2.54	1.27	
Iliotibial band strain rate (%/s)	42.87	14.67	42.96	12.71	30.30	7.37	28.32	8.06	B
Patellofemoral integral (BW·s)	0.29	0.08	0.26	0.08	0.31	0.13	0.28	0.10	A
Patellofemoral loading rate (BW/s)	156.50	55.49	154.50	33.75	179.18	48.37	143.14	22.99	
Patellofemoral stress integral (KPa/BW·s)	0.56	0.14	0.52	0.13	0.59	0.21	0.55	0.18	
Patellofemoral stress loading rate (KPa/BW/s)	323.28	125.29	326.40	81.24	375.92	117.61	302.65	59.30	
Achilles integral (BW·s)	0.65	0.07	0.71	0.05	0.71	0.05	0.79	0.12	A, B
Achilles loading rate (BW/s)	153.96	43.34	179.34	67.96	179.34	67.96	148.14	38.75	
Ankle integral (BW·s)	1.21	0.12	1.30	0.12	1.30	0.12	1.33	0.19	
Ankle loading rate (BW/s)	251.82	41.42	281.18	55.95	281.18	55.95	247.14	40.27	
Hip integral (BW·s)	1.34	0.16	1.31	0.09	1.31	0.09	1.25	0.13	
Hip loading rate (BW/s)	276.22	41.21	291.88	82.86	291.88	82.86	260.00	123.79	
Medial tibiofemoral integral (BW·s)	0.86	0.10	0.83	0.06	0.83	0.06	0.85	0.12	
Medial tibiofemoral loading rate (BW/s)	212.57	51.75	274.96	75.23	274.96	75.23	196.17	64.60	C
Lateral tibiofemoral integral (BW·s)	0.44	0.07	0.44	0.05	0.44	0.05	0.41	0.07	
Lateral tibiofemoral loading rate (BW/s)	157.20	63.56	151.07	38.23	151.07	38.23	130.20	36.26	
Strike index (%)	22.61	17.92	33.79	24.69	46.48	21.44	61.68	19.33	A, B

A = main effect of FOOTWEAR
B = main effect of GROUP
C = FOOTWEAR x GROUP interaction

648 Table 2: Discrete temporal biomechanical parameters (mean ± standard deviations) as a function of
649 FOOTWEAR and GROUP.

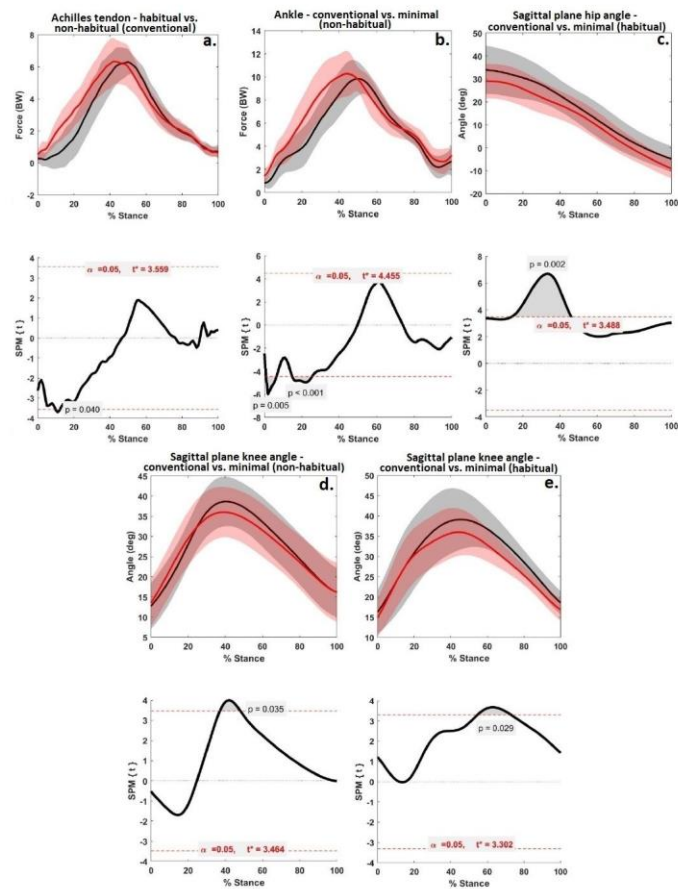| | Non-habitual | | | | Habitual | | | | |
| | Conventional | | Minimal | | Conventional | | Minimal | | |
	Mean	SD	Mean	SD	Mean	SD	Mean	SD	
Step length (m)	1.41	0.14	1.4	0.15	1.29	0.17	1.23	0.15	A, B, C
Patellofemoral integral per kilometre m (BW·km)	543.91	163.20	493.74	160.51	646.21	300.30	612.77	253.39	
Patellofemoral stress integral per kilometre (KPa/BW·km)	1048.04	280.21	970.31	268.78	1203.71	494.77	1188.42	439.10	
Achilles integral per kilometre (BW·km)	1196.94	174.21	1328.26	134.48	1446.36	181.08	1697.98	361.64	A, B
Ankle integral per kilometre (BW·km)	2255.40	334.19	2410.35	249.97	2637.91	428.70	2849.76	582.01	B
Hip integral per kilometre (BW·km)	2507.87	504.79	2465.54	403.70	2672.95	391.66	2676.06	452.34	
Medial tibiofemoral integral per kilometre (BW·km)	1608.00	298.37	1561.71	234.69	1694.68	239.98	1826.23	378.21	
Lateral tibiofemoral integral per kilometre (BW·km)	815.75	191.45	826.17	133.47	902.91	173.35	875.81	201.87	

650   A = main effect of FOOTWEAR
651   B = main effect of GROUP
652   C = FOOTWEAR x GROUP interaction

653

654

655
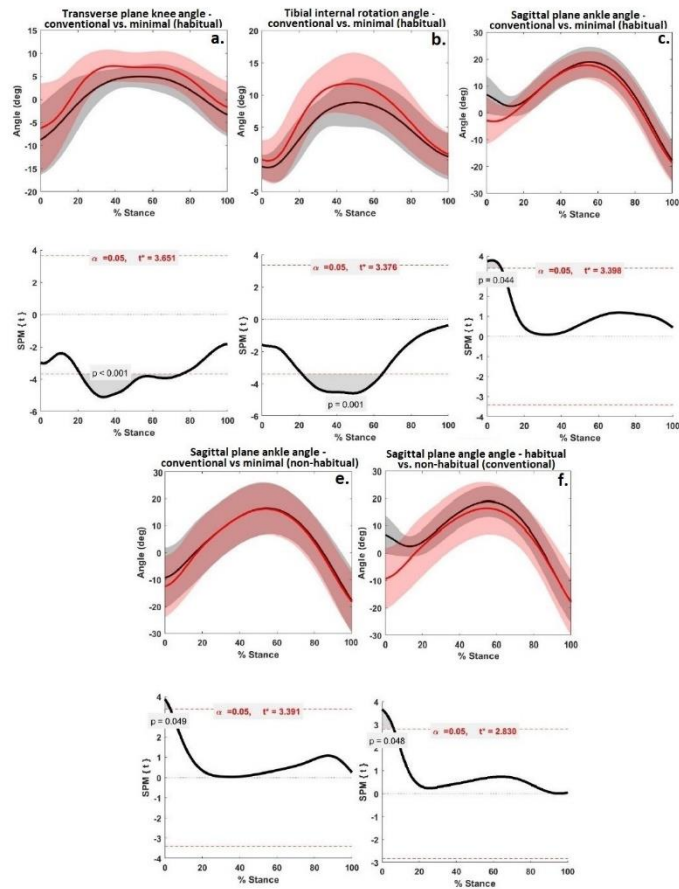
656

657

658

659

660

661

662

663

664

665

666

667

668 **Figure labels**

669

670 Figure 1: Experimental footwear (A = conventional and B = minimal).

671

Figure 2: Statistical parametric mapping results of Achilles tendon and ankle forces in addition to hip and knee kinematics (FOOTWEAR: black = conventional/ red = minimal & GROUP: black = non-habitual/ red = habitual).

675

Figure 3: Statistical parametric mapping results of tibial internal rotation, knee and ankle kinematics (FOOTWEAR: black = conventional/ red = minimal & GROUP: black = non-habitual/ red = habitual).