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Acute biomechanical effects of a lightweight, sock-style minimalist footwear design during running; a musculoskeletal simulation and statistical parametric mapping approach

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1 **Acute biomechanical effects of a lightweight, sock-style minimalist footwear design**
2 **during running; a musculoskeletal simulation and statistical parametric mapping**
3 **approach.**

4
5 **Abstract**

6 The aim of this study was to examine the effects of existing minimalist footwear, new sock-
7 style minimalist footwear and conventional running footwear on lower extremity
8 biomechanics, using a musculoskeletal simulation and statistical parametric mapping (SPM)
9 approach. Thirteen male participants ran over an embedded force plate at 4.0 m/s, in 1.
10 existing minimalist footwear, 2. new sock-style minimalist footwear and 3. conventional
11 running shoes. Kinematics of the lower extremities were collected using an eight-camera
12 motion analysis system and lower extremity joint loading was also explored using a
13 musculoskeletal simulation approach. Differences between footwear conditions were
14 examined using SPM and one-way repeated measures ANOVA. The strike index indicated
15 that the foot contact position was significantly more anterior in existing minimalist footwear
16 (44.19 %) and new sock-style minimalist footwear (42.33 %) compared to conventional
17 running shoes (29.00 %). The instantaneous loading rate was also significantly larger in
18 existing minimalist footwear (271.68 BW/s) and new sock-style minimalist footwear (299.26
19 BW/s) in relation to conventional running shoes (122.48 BW/s). In addition, during the late
20 stance phase compressive hip joint loading was significantly larger in both minimalist
21 footwear. Similarly, Achilles tendon loading was statistically greater in both minimalist
22 footwear compared to the conventional running shoe during the early and middle aspects of
23 the stance phase. The observations from this analysis show that minimalist footwear may

24 place non-habituated runners at greater risk from the mechanical factors linked to the
25 aetiology of chronic lower limb running related injuries.

26

27 Introduction

28 Running is one of the most popular aerobic exercise modalities, and there is an overwhelming
29 body of evidence that it mediates a plethora of physiological and psychological benefits (Lee
30 et al., 2014). However, running is also associated with an extremely high susceptibility to
31 chronic pathologies; with up to 80 % of runners experiencing an injury each year (Van Gent
32 et al., 2007). Chronic injuries are a key barrier to training compliance (Hespanhol et al.,
33 2016), and result in a significant economic burden due to healthcare operation and absence
34 from work (Junior et al., 2017).

35

36 As the primary interface between foot and ground, running shoes are proposed as a
37 mechanism by which the rate of chronic injuries can be moderated (Shorten, 2000). However,
38 since the introduction of the conventional running shoe in the 1970's, the rate and location of
39 chronic running injuries has remained unchanged (Davis, 2014). This has led to the
40 supposition that reverting to running in minimalist footwear that lacks the mechanical
41 properties associated with the conventional running shoe, may be associated with a reduced
42 incidence of chronic running injuries (Lieberman et al., 2010). Based on this supposition
43 several minimalist footwear models such as the Vibram Five-Fingers are currently available
44 commercially.

45

46 Several studies have explored biomechanical differences between minimalist and
47 conventional running shoes. These analyses have typically examined spatiotemporal
48 characteristics, lower limb kinematics and loading rates. Sinclair et al., (2013a) and Sinclair
49 et al., (2016) showed that minimalist footwear caused runners to run with a more
50 plantarflexed ankle at initial contact, increased peak tibial internal rotation and an increased
51 vertical loading rate in comparison to conventional running shoes. Squadrone et al., (2009)
52 similarly showed that running in minimalist footwear increased the ankle plantarflexion angle
53 at footstrike but also reduced stride length and the impact peak of the vertical ground reaction
54 force (GRF). Squadrone et al., (2015) investigated the effects of different minimalist
55 footwear conditions via the strike index. Their findings showed that minimalist footwear
56 mediated a midfoot strike pattern, with alterations being most pronounced in footwear with
57 the least midsole cushioning. Sinclair et al., (2018) showed that the strike index did not
58 change between different minimalist footwear models and conventional running shoes, but
59 did find that effective mass was significantly larger in minimalist footwear with alterations
60 again being more evident in models with the least midsole cushioning.

61

62 Previous work has also examined the effects of minimalist footwear on the loads experienced
63 by the lower extremities joint during running. Sinclair, (2014) and Sinclair et al., (2016)
64 showed that peak patellofemoral stress was significantly reduced in minimalist footwear, but
65 peak Achilles tendon loads were significantly increased. Similarly, Bonacci et al., (2018)
66 showed that peak patellofemoral stress was significantly lower in minimalist footwear. In
67 addition, Sinclair, (2016) showed that peak tibiofemoral loading did not differ significantly
68 between minimalist and conventional footwear during running. Furthermore, Sinclair et al.,
69 (2015) and Sinclair et al., (2016) taking into account the effect of changing stride length
70 examined the effects of different minimalist footwear. Patellofemoral impulse per mile was

71 significantly reduced but Achilles tendon impulse per mile was significantly greater in
72 minimalist footwear, with differences being more evident in minimalist footwear with the
73 least midsole cushioning. Recently, a new lightweight, sock-style minimalist footwear design
74 has been commercially released, which represents an extremely lightweight sock style upper
75 with a strong abrasion resistant sole. There are however, no published scientific
76 investigations concerning this new footwear, indicating that examination of running
77 biomechanics whilst wearing these shoes is warranted.

78

79 Previous analyses concerning the biomechanical differences between minimalist and
80 conventional footwear, have utilized mathematical modelling approaches driven by joint
81 torques to explore the loads experienced by the musculoskeletal system. However, joint
82 torques are global indices of joint loading, and therefore not representative of localized joint
83 loading (Herzog et al., 2003a). Herzog et al., (2003b) identified importantly that the muscles
84 are the primary contributors to lower extremity joint loading. Due to the difficulties
85 associated with calculating muscle kinetics, the role of the muscles in controlling joint
86 biomechanics during running has received little attention within biomechanical literature.
87 Over the past decade however, significant advances have been made in improving
88 musculoskeletal models; leading to the development of open access and bespoke software.
89 Allowing skeletal muscle forces to be simulated during movement, and utilized as inputs to
90 calculate lower extremity joint reaction forces (Delp et al., 2007). Such approaches have not
91 yet been utilized to explore biomechanical differences between minimalist and conventional
92 running shoes.

93

94 To date biomechanical differences between minimalist and conventional footwear have been
95 explored statistically through extraction of discrete kinetic/ kinematic parameters. This
96 approach can however be limiting, as it can lead to potentially relevant information being
97 discarded (Warmenhoven et al., 2018). Therefore, Statistical parametric mapping (SPM) may
98 represent an efficacious supplement to discrete analyses, as it is able to compare an entire
99 time normalized data series (Pataky et al., 2013). To date there has yet to be any
100 biomechanical investigation, which has examined the effects of different minimalist footwear
101 and conventional running shoes on the biomechanical parameters linked to the aetiology of
102 running injuries using SPM.

103

104 To summarize, there is currently no scientific research concerning the aforementioned sock-
105 style minimalist footwear, nor is there any investigation which has collectively explored the
106 effects of minimalist and conventional running shoes using both musculoskeletal simulation
107 and SPM. Therefore, the aim of the current investigation was to examine the effects of
108 existing/ sock-style minimalist footwear and conventional running shoes on lower extremity
109 biomechanics using a musculoskeletal simulation and SPM based approach. A study of this
110 nature may provide further insight into the biomechanical differences between minimalist and
111 traditional running shoes; particularly with regards to runners' predisposition to chronic
112 running injuries.

113

114 **Methods**

115 *Participants*

116 Thirteen male runners volunteered to take part in this study. This sample size is
117 commensurate with previous analyses concerning the biomechanics of running in minimalist

118 footwear (Sinclair et al., 2013a; Sinclair et al., 2015). The mean characteristics of the
119 participants were: age 27.31 ± 3.50 years, height 1.73 ± 0.04 m and body mass 72.23 ± 5.66
120 kg. The procedure utilized for this investigation was approved by the University of Central
121 Lancashire, Science, Technology, Engineering and Mathematics, ethical committee. All
122 runners were free from musculoskeletal pathology at the time of data collection. Participants
123 provided written informed consent in accordance with the principles outlined in the
124 Declaration of Helsinki.

125

126 *Footwear*

127 The footwear used during this study consisted of New Balance, 1260 v2 (New Balance,
128 Boston, Massachusetts, United States; henceforth termed Shoe A), Vibram Five-Fingers,
129 ELX (Vibram, Albizzate, Italy; henceforth termed Shoe B) and Skinnners, Athleisure
130 (Skinnners Technologies, Cyrilska, Czech Republic; henceforth termed Shoe C) (Figure 1).
131 Shoe A had an average mass of 0.285 kg, heel thickness of 25 mm and a heel drop of 14 mm.
132 Shoe B had an average mass of 0.167 kg, heel thickness of 7 mm and a heel drop of 0 mm.
133 Finally, Shoe C had an average mass of 0.08 kg, heel thickness of 6 mm and a heel drop of 0
134 mm. The footwear were also scored using the minimalist index described by Esculier et al.,
135 (2015), and Shoe A received a score of 20, Shoe B a score of 92 and Shoe C a score of 100.

136

137 @@@ *Figure 1 near here* @@@

138

139 *Procedure*

140 Participants ran at 4.0 m/s ($\pm 5\%$), striking an embedded piezoelectric force platform (Kistler
141 Instruments Ltd., Winterthur, Switzerland) with their right foot. Running velocity was
142 monitored using infrared timing gates (Newtest, Oy Kouluatu, Finland). The stance phase
143 was delineated as the duration over which 20 N or greater of vertical GRF was applied to the
144 force platform. Runners completed a minimum of five successful trials in each footwear
145 condition. As each footwear were novel to all participants, a period of 5 minutes for
146 accommodation was allowed. This involved running through the testing area without concern
147 for striking the force platform (Sinclair et al., 2013a; Sinclair et al., 2016). The order that
148 participants ran in each footwear condition was counterbalanced. Kinematic and GRF data
149 were synchronously collected. Kinematic data were captured at 250 Hz via an eight-camera
150 motion analysis system (Qualisys Medical AB, Goteburg, Sweden). Dynamic calibration of
151 the motion capture system was performed before each data collection session.

152

153 Lower extremity segments were modelled in 6 degrees of freedom using the calibrated
154 anatomical systems technique (Cappozzo et al., 1995). To define the anatomical frames of the
155 thorax, pelvis, thighs, shanks and feet retroreflective markers were placed at the C7, T12 and
156 xiphoid process landmarks and also positioned bilaterally onto the acromion process, iliac
157 crest, anterior superior iliac spine (ASIS), posterior superior iliac spine (PSIS), medial and
158 lateral malleoli, medial and lateral femoral epicondyles, greater trochanter, calcaneus, first
159 metatarsal and fifth metatarsal. Carbon-fibre tracking clusters comprising of four non-linear
160 retroreflective markers were positioned onto the thigh and shank segments. In addition to
161 these, the foot segments were tracked via the calcaneus, first metatarsal and fifth metatarsal,
162 the pelvic segment was tracked using the PSIS and ASIS markers and the thorax segment was
163 tracked using the T12, C7 and xiphoid markers. Static calibration trials (not normalized to

164 static trial posture) were obtained in each footwear allowing for the anatomical markers to be
165 referenced in relation to the tracking markers/ clusters.

166

167 *Processing*

168 Dynamic trials were digitized using Qualisys Track Manager (Qualisys Medical AB,
169 Goteburg, Sweden) in order to identify anatomical and tracking markers then exported as
170 C3D files to Visual 3D (C-Motion, Germantown, MD, USA). All data were linearly
171 normalized to 100 % of the stance phase. GRF data and marker trajectories were smoothed
172 with cut-off frequencies of 50 Hz at 12 Hz respectively, using a low-pass Butterworth 4th
173 order zero lag filter. All force parameters throughout were normalized by dividing by
174 bodyweight (BW).

175

176 In accordance with the protocol of Addison & Lieberman, (2015), an impulse-momentum
177 modelling approach was utilized to calculate effective mass (% BW), which was quantified in
178 accordance with the below equation:

179

$$180 \text{ Effective mass} = \text{vertical GRF integral} / (\Delta \text{foot vertical velocity} + g * \Delta \text{time})$$

181

182 The impact peak was defined in Shoe A as the first peak in vertical GRF. In Shoes B and C
183 where no impact peak was present, according to the protocols of Lieberman et al., (2010) and
184 Sinclair et al., (2018) we defined the position of the impact peak at the same relative position
185 as in Shoe A, which was shown to be 11.96 % of the stance phase. The time (ms) to impact
186 peak (Δtime) was quantified as the duration from footstrike to impact peak. The vertical GRF

187 integral (BW·ms) during the period of the impact peak was calculated using a trapezoidal
188 function. The change in foot **vertical** velocity (Δ foot **vertical** velocity) was determined as the
189 instantaneous vertical foot velocity averaged across the 10 frames prior to the impact peak
190 **(Sinclair et al., (2018))**. The velocity of the foot was quantified using the centre of mass of the
191 foot segment in the vertical direction, within Visual 3D (Sinclair et al., 2018).

192

193 **Instantaneous loading rate (BW/s) was also was also extracted by obtaining the peak increase**
194 **in vertical GRF between adjacent data points.** Finally, the strike index was calculated as the
195 position of the centre of pressure location at **footstrike**, relative to the total length of the foot
196 (Squadrone et al., 2015). A strike index of 0–33% denotes a rearfoot, 34–67% a midfoot and
197 68–100% a forefoot strike pattern.

198

199 Following this, data during the stance phase were exported from Visual 3D into OpenSim 3.3
200 software (Simtk.org). A validated musculoskeletal model with 12 segments, 19 degrees of
201 freedom and 92 musculotendon actuators (Lerner et al., 2015) was used to estimate lower
202 extremity joint forces. The model was scaled to account for the anthropometrics of each
203 athlete. As muscle forces are the main determinant of joint compressive forces (Herzog et al.,
204 2003), muscle kinetics were quantified using static optimization in accordance with Steele et
205 al., (2012). Compressive patellofemoral, medial/ lateral tibiofemoral and hip joint forces were
206 calculated via the joint reaction analyses function using the muscle forces generated from the
207 static optimization process as inputs. **Finally, Achilles tendon forces were estimated in**
208 **accordance with the protocol of Almonroeder et al., (2013), by summing the muscle forces of**
209 **the medial gastrocnemius, lateral, gastrocnemius, and soleus muscles.**

210

211 Running in minimalist footwear has been shown to alter step length during running (Sinclair
212 et al., 2016), which increases the number of footsteps necessary to run a set distance. We
213 therefore firstly calculated hip, tibiofemoral, patellofemoral and Achilles tendon impulse
214 during the stance phase, using a trapezoidal function. In addition to this, we also estimated
215 the total impulse per kilometre (BW·km) by multiplying these parameters by the number of
216 steps required to run a kilometre. The number of steps required to complete one kilometre
217 was quantified using the step length (m), which was determined by taking the difference in
218 the horizontal position of the foot centre of mass between the right and left legs at footstrike.

219

220 *Statistical analyses*

221 Compressive joint forces (hip, patellofemoral, medial tibiofemoral and lateral tibiofemoral),
222 Achilles tendon loading and three-dimensional kinematics during the entire stance phase
223 were temporally normalized using linear interpolation to 101 data points. Differences across
224 the entire stance phase were examined using 1-dimensional SPM with MATLAB 2017a
225 (MATLAB, MathWorks, Natick, USA), in accordance with Pataky et al., (2016), using the
226 source code available at <http://www.spm1d.org/>. In agreement with Pataky et al., (Pataky et
227 al., 2013), SPM was implemented in a hierarchical manner, analogous to one-way repeated
228 measures ANOVA (SPM F) with post-hoc paired t-tests (SPM t). Therefore, the entire data
229 set was examined first, and if a statistical main effect was reached, then post-hoc tests were
230 conducted on each component separately.

231

232 For discrete parameters that could not be examined using SPM (hip impulse per km, lateral
233 impulse per km, medial impulse per km, patellofemoral impulse per km, Achilles tendon
234 impulse per km, step length, instantaneous load rate, strike index and effective mass), means

235 and standard deviations were calculated for each outcome measurement for all footwear
236 conditions. Differences in discrete biomechanical parameters between footwear were
237 examined using one-way repeated measures ANOVAs, Effect sizes were calculated using
238 partial eta² (η^2). In the event of a significant main effect, post-hoc pairwise comparisons
239 were conducted on all significant main effects, using a Bonferroni adjustment. Discrete
240 statistical actions were conducted using SPSS v24.0 (SPSS Inc., Chicago, USA). Statistical
241 significance for main effects was accepted at the $P \leq 0.05$ level (Sinclair et al., 2013b).

242

243 **Results**

244 @@@ *Figure 2 near here* @@@

245 @@@ *Figure 3 near here* @@@

246 @@@ *Table 1 near here* @@@

247

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

249 A main effect was revealed for the instantaneous **loading** rate ($P < 0.001$, $\eta^2 = 0.75$). Post-hoc
250 analyses showed that instantaneous **loading** rate was significantly larger in Shoe B ($P < 0.001$)
251 and Shoe C ($P < 0.001$), compared to Shoe A (Table 1).

252

253 A main effect was shown for strike index ($P = 0.033$, $\eta^2 = 0.27$). Post-hoc analyses showed
254 that strike index was significantly larger in Shoe B ($P = 0.008$) and Shoe C ($P = 0.006$),
255 compared to Shoe A (Table 1).

256

257 A main effect was evident for effective mass ($P=0.005$, $\eta^2 = 0.38$). Post-hoc analyses
258 showed that effective mass was significantly larger in Shoes A ($P=0.01$) and C ($P=0.04$),
259 compared to Shoe B (Table 1). Finally, a main effect was shown for step length ($P=0.012$,
260 $\eta^2 = 0.33$). Post-hoc analyses showed that step length was significantly larger in Shoe A
261 compared to Shoe C ($P=0.005$) (Table 1).

262

263 *Joint loading per kilometre*

264 At the hip joint a main effect was found for peak hip impulse per kilometre ($P=0.018$, $\eta^2 =$
265 0.31). Post-hoc analysis showed that hip impulse per kilometre was significantly larger in
266 Shoe C compared to shoe A ($P=0.004$) (Table 1).

267

268 There was also a main effect for patellofemoral impulse per kilometre ($P=0.029$, $\eta^2 = 0.28$).
269 Post-hoc analysis showed that patellofemoral impulse per kilometre was significantly larger
270 in Shoe C compared to shoe B ($P=0.02$) (Table 1).

271

272 Finally, a main effect was found for Achilles tendon impulse per kilometre ($P<0.001$, $\eta^2 =$
273 0.58). Post-hoc analyses showed that Achilles tendon impulse per kilometre was significantly
274 larger in Shoes B ($P=0.001$) and C ($P=0.002$) compared to shoe A (Table 1).

275

276 *Statistical parametric mapping - joint loading*

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278

@@@ *Figure 4 near here* @@@

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@@@ *Figure 5 near here* @@@

@@@ *Figure 6 near here* @@@

At the hip joint, there was a significant main effect (Figure 4a). Post-hoc analyses showed that Shoe A was associated with lower compressive hip force than Shoes B and C, from 82-88% of the stance phase (Figure 4bc).

At the patellofemoral joint, there was a significant main effect (Figure 4d). Post-hoc analyses showed that Shoe A was associated with lower patellofemoral force than Shoe B from 81-90% of the stance phase (Figure 4e).

At the medial aspect of the tibiofemoral joint, there was also a main effect (Figure 4f). Post-hoc analyses showed that Shoe A was associated with lower compressive force than Shoe B from 5-10% and 80-92% of the stance phase (Figure 4g). In addition, Shoe A was associated with lower compressive loading than Shoe C from 5-10% of the stance phase yet greater loading from 4-9% of the stance phase (Figure 4h).

At the lateral aspect of the tibiofemoral joint, there was also a main effect (Figure 5a). Post-hoc analyses showed that Shoe A was associated with lower compressive force than Shoe B 82-89% of the stance phase (Figure 5b). In addition, Shoe A was associated with lower compressive force than Shoe C, between 0-3% of the stance phase (Figure 5c).

301 At the Achilles tendon, there was a main effect (Figure 5d). Post-hoc analyses showed that
302 Shoe A was associated with lower tendon loading than Shoe B, between 7-12%, 17-55% and
303 82-92% of the stance phase (Figure 5e). In addition, Shoe A was associated with lower
304 tendon loading compared to Shoe C, from 0-3%, 20-25% and 35-50% of the stance phase
305 (Figure 5f).

306

307 *Statistical parametric mapping - three-dimensional kinematics*

308 For tibial internal rotation, there was a main effect (Figure 5g). Post-hoc analyses showed that
309 Shoe A was associated with **increased** tibial internal rotation than Shoe B, between 0-5% and
310 90-100% of the stance phase (Figure 5h).

311

312 At the ankle in the sagittal plane, there was a main effect (Figure 6a). Post-hoc analyses
313 showed that Shoe A was significantly more dorsiflexed than Shoe B, from 0-3% of the stance
314 phase (Figure 6b). In addition, it was revealed that Shoe A was significantly more dorsiflexed
315 than Shoe C, from 0-8% of the stance phase (Figure 6c).

316

317 **Discussion**

318 **The current investigation aimed to examine the effects of existing/ sock-style minimalist**
319 **footwear and conventional running shoes on lower extremity biomechanics using a**
320 **musculoskeletal simulation and SPM based approach. To the authors knowledge this is the**
321 **first investigation to comparatively examine these footwear and to explore the biomechanics**
322 **of running in conventional and minimalist footwear using musculoskeletal simulation and**
323 **SPM.**

324

325 The kinematic analysis using SPM showed that the ankle was in a significantly more
326 plantarflexed position during the early stance phase in Shoes B and C in comparison to Shoe
327 A. This observation is reinforced by the discrete point analysis of the strike index, which
328 showed that the contact position was significantly more anterior in Shoes B and C, and a
329 midfoot strike pattern was adopted when wearing these footwear. This finding concurs with
330 the observations of Sinclair et al., (2013a) and Sinclair et al., (2016) who each showed an
331 altered foot position when wearing minimalist footwear. It is proposed that this relates to the
332 absence of cushioning in Shoes B and C, causing runners to adopt a flatter foot position in
333 order to compensate for the lack of midsole interface in an attempt to attenuate the load
334 experienced by the lower extremities (Lieberman et al., 2010).

335

336 The findings from the current investigation also showed that the instantaneous loading rate
337 was significantly larger and the effective mass was significantly lower in Shoes B and C
338 compared to Shoe A. This observation agrees with those of Sinclair et al., (2013a) and
339 Sinclair et al., (2016) but opposes those of Squadrone & Gallozzi, (2009) and Sinclair et al.,
340 (2018). Transient loading is governed by the rate at which the momentum of the foot
341 changes, therefore midsole material at the foot-ground interface strongly influences the
342 magnitude of transient forces during running (Whittle, 1999). Importantly, Addison &
343 Liebermann, (2015) found that the loading rate and effective mass were inversely associated
344 during running. Therefore, the aforementioned observation in relation to the loading rate is
345 supported by the effective mass observations, which was shown to be reduced in Shoes B and
346 C compared to Shoe A. Given the proposed association between the instantaneous rate of
347 loading and the aetiology of chronic injuries, this finding may be clinically meaningful,

348 (Milner et al., 2006), and indicates that Shoes B and C may place runners at increased risk
349 from impact related injuries compared to Shoe A.

350

351 At the hip joint, the current investigation showed using SPM, that Shoe A significantly
352 reduced compressive hip joint loading during the early and late aspects of the stance phase
353 compared to Shoes B and C. This observation is supported through the discrete point
354 analysis, which showed that compressive joint forces experienced per kilometre were
355 statistically greater in Shoe C compared to shoe A. As the current investigation represents the
356 first investigation to compare hip joint loading when running in minimalist and conventional
357 footwear using musculoskeletal simulation, comparisons in relation to previous analyses are
358 not possible. Nonetheless, the results are partially supported by those of Rooney & Derrick,
359 (2013) and Sinclair, (2018) who showed that modifying the foot position significantly
360 enhanced compressive hip joint loading during running. As the aetiology of hip joint
361 pathologies are strongly influenced by compressive hip joint loading (Johnson & Hunter,
362 2014), the current investigation indicates that Shoes B and C may increase runners'
363 susceptibility to chronic hip pathologies.

364

365 A further important observation from the current analysis is that patellofemoral loading
366 contrasted using SPM was statistically larger in Shoe B compared to Shoe A during late
367 stance. The discrete analysis differed from this, showing that patellofemoral force per
368 kilometre was significantly larger in Shoe C compared to shoe B. The observations from the
369 current investigation oppose those of Sinclair, (2014), Sinclair et al., (2016) and Bonacci et
370 al., (2018) who showed significant reductions in peak patellofemoral stress and
371 patellofemoral impulse per mile when running in minimalist footwear. This observation may

372 be due to the mechanism by which patellofemoral forces were calculated, as previous utilized
373 mathematical models have not accounted for muscular co-contraction, and Sinclair, (2018)
374 similarly showed using musculoskeletal simulation that running barefoot did not attenuate
375 patellofemoral kinetics compared to conventional running shoes. The current investigation
376 indicates firstly that running in minimalist footwear may not necessarily attenuate the
377 magnitude of patellofemoral loading linked to the aetiology of patellofemoral disorders
378 during running, in relation to conventional running shoes. Furthermore, the current study
379 revealed that patellofemoral was statistically larger in Shoe C compared to shoe B, indicating
380 that despite their relatively similar design characteristics (Esculier et al., 2015); Shoe C may
381 place runners at increased risk from patellofemoral chronic injuries.

382

383 At the medial and lateral tibiofemoral joint compartments, compressive loading was
384 significantly greater in Shoes B and C in relation to Shoe A, during the early and late aspects
385 of the stance phase. This observation opposes those of Sinclair, (2016) but is supported
386 closely by those of Sinclair, (2018); who showed that the medial and lateral tibiofemoral
387 compressive rate of loading was statistically greater when running barefoot. This observation
388 may be clinically meaningful, as increased compressive loading at both aspects of the
389 tibiofemoral joint, is recognised as the primary risk factor in relation to the aetiology and
390 progression of osteoarthritic symptoms (Dabiri & Li, 2013). Therefore, the current study
391 shows that indicates that running in minimalist footwear may increase runners predisposition
392 to the risk factors linked to the initiation of tibiofemoral osteoarthritis.

393

394 The findings from the current investigation also revealed using SPM that Achilles tendon
395 loading was statistically larger during the mid and late aspects of the stance phase in Shoes B

396 and C compared to Shoe A. In addition, the discrete point analysis of tendon loading per
397 kilometre similarly indicated that Shoes B and C were associated with statistically larger
398 tendon loading magnitudes. This observation concurs with those of Sinclair, (2014) and
399 Sinclair et al., (2015) who similarly showed that peak Achilles tendon force and tendon
400 impulse per mile were greater when running in minimalist footwear in comparison to
401 conventional running shoes. The aetiology of Achilles tendinopathy is associated with
402 excessive and repeated tendinous loading, during cyclic activities such as running
403 (Magnusson et al., 2010). Excessive tendon loading without sufficient caseation of running
404 activities between training sessions, mediates collagen and extracellular matrix synthesis and
405 degradation of the tendon (Magnusson et al., 2010). As such, the current investigation shows
406 that running in minimalist footwear may place runners at increased risk from the
407 biomechanical parameters linked to Achilles tendinopathy, in comparison to conventional
408 running shoes.

409

410 A potential limitation that should be acknowledged in regards to the current investigation is
411 of course that only runners who habitually ran in conventional running shoes were examined.
412 The findings from previous analyses concerning the biomechanics of minimalist footwear
413 and conventional running shoes have drawn opposing interpretations, frequently on the basis
414 of the running experience of the participants in minimalist footwear (Sinclair et al., 2013a;
415 Squadrone & Gallozzi, 2009). It can therefore be ventured that the findings from the current
416 investigation may have been different, had the participants been habitual minimalist footwear
417 users. As such, future analyses using musculoskeletal simulation and SPM investigating the
418 biomechanics of running in habitual minimalist footwear is recommended, allowing more
419 decisive assertions in regards to the aetiology of chronic pathologies to be drawn.

420

421 In conclusion, though the biomechanics of running in minimalist and conventional running
422 footwear have received widespread research attention, there has not yet been a quantitative
423 comparison of lower extremity biomechanics in minimalist and conventional running shoes
424 using a musculoskeletal simulation and SPM based approach. This study revealed that the
425 instantaneous load rate, hip, tibiofemoral and Achilles tendon force parameters were
426 statistically larger when running in Shoes B and C compared to Shoe A. Therefore, the
427 observations from this analysis show that minimalist footwear may place non-habituated
428 runners at greater risk from the mechanical factors linked to the aetiology of chronic lower
429 limb running related injuries.

430

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536 **Figure 1: Experimental footwear (A = New Balance/ Shoe A, B = Vibram Five-Fingers/ Shoe
537 B and C = Skidders/ Shoe C).**

538 Figure 2: Hip, knee and ankle kinematics in the a. sagittal, b. coronal and c. transverse planes
539 as a function of footwear (black = Shoe A, dash = Shoe B and grey = Shoe C), (FL = flexion,
540 AD = adduction, IN = inversion, INT = internal, EXT = external).

541 Figure 3: Lower extremity joint loading as a function of footwear (black = Shoe A, dash =
542 Shoe B and grey = Shoe C), (a. = hip, b. = patellofemoral, c. = medial tibiofemoral, d. =
543 lateral tibiofemoral and e. Achilles tendon).

544 Figure 4: Statistical parametric mapping results in relation to lower extremity joint loading (a.
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549 Figure 5: Statistical parametric mapping results in relation to lower extremity joint loading
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