

1 **Effects of different heel heights on lower extremity joint loading in experienced and in-**  
2 **experienced users: A musculoskeletal simulation analysis.**

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21

## 22 **Abstract**

23 **Purpose:** This study examined the effects of different high-heeled footwear heights on lower  
24 extremity compressive joint loading and triceps-surae muscle tendon kinematics during  
25 walking, using a musculoskeletal simulation based approach, in both experienced and in-  
26 experienced high heel users.

27 **Methods:** The current investigation examined 12 experienced and 12 inexperienced high-  
28 heel wearers, walking in four different footwear (high heel, medium heel, low heel and  
29 trainer). Walking kinematics were collected using an eight-camera motion capture system,  
30 and kinetics via an embedded force plate. Lower extremity joint loading and triceps-surae  
31 muscle kinematics were explored using a musculoskeletal simulation approach.

32 **Results:** Irrespective of experience, when wearing high-heels of increasing height,  
33 compressive loading parameters at the medial tibiofemoral compartment and patellofemoral  
34 joint were significantly greater and exceeded the minimum clinically important difference  
35 (MCID). Furthermore, irrespective of wearers' experience, the triceps-surae muscle tendon  
36 units were placed in a shortened position when wearing high-heels of increasing height, with  
37 the differences exceeding the MCID.

38 **Conclusions:** It can be concluded that heeled-footwear increase the mechanical factors linked  
39 to the aetiology of degenerative joint osteoarthritis, and chronic shortening of the triceps-  
40 surae muscle tendon units. Therefore, the current investigation provides evidence that  
41 irrespective of experience, heeled-footwear of increasing height may negatively influence  
42 female's lower extremity musculoskeletal health.

43

## 44 **Introduction**

45 Walking is a fundamental aspect of everyday living, and the principal locomotion modality in  
46 humans. High-heeled shoes have been a prevalent footwear choice for over 400 years and are  
47 used daily in 39-69% of females (1). Heeled designs remain one of the central features of  
48 women's footwear, and social and fashion practices promote the continued use of high-heels  
49 (2). Although millions of women wear heeled footwear, concerns regarding the chronic  
50 impact of high-heels on women's musculoskeletal health have been articulated for over 50  
51 years (1).

52

53 High-heeled footwear feature a slender base of support, and force the ankle in to a plantar-  
54 flexed state, mediating kinematic and kinetic changes in lower extremity biomechanics  
55 during walking (3). A substantial literature base currently exists concerning the biomechanics  
56 of walking in heeled footwear. Stefanyshyn et al., (4) examined heel heights of 1.4, 3.7, 5.4  
57 and 8.5cm and showed firstly that peak braking/ propulsive forces, in addition to the active  
58 peak of the vertical ground reaction force, increased linearly. This study also showed graded  
59 increases in knee/ ankle flexion and activation of the rectus femoris and soleus musculature.  
60 Naik et al., (5) examined 4, 6, 8, 10, and 12cm heel heights during a stand-to-sit-returning  
61 task. Their findings showed imbalances between vastus lateralis and medialis muscles that  
62 became more prominent in elevated heels. Simonsen et al., (6), Esenyel et al., (7) and  
63 Kerrigan et al., (8) each found that the magnitude of knee extensor moment during the first  
64 half of the stance phase was substantially larger, which was attributed to increased knee  
65 flexion when wearing heeled footwear. In addition, several investigations have shown that the  
66 magnitude of the external knee adduction moment increased significantly in high-heeled  
67 footwear compared to flat shoes and also linearly with increases in heel height (6, 9, 10). At  
68 the ankle joint, Barkema et al., (9) found that the peak ankle eversion moment during late  
69 stance phase was amplified linearly with increases in heel height. Both, Esenyel et al., (7) and

70 Simonsen et al., (6) showed that the peak plantarflexion moment at the end of the stance  
71 phase was significantly reduced in heeled footwear. Finally, at the hip joint Simonsen et al.,  
72 (6) showed that the hip joint abductor moment was significantly larger when walking in high-  
73 heels.

74

75 Though females regularly wear high-heeled shoes, it has been suggested that their continued  
76 utilization may lead to an increased incidence of chronic musculoskeletal pathologies.  
77 Importantly, previous analyses have shown that stance phase knee adduction moments were  
78 statistically larger when walking in high-heels (6, 9, 10). Leading to the proposition that high-  
79 heels may augment compressive knee joint loading, placing wearers at risk from tibiofemoral  
80 joint osteoarthritis. A musculoskeletal condition renowned for its increased prevalence in  
81 females (11). Importantly, biomechanical accommodations to high-heels have been shown to  
82 vary with experience in wearing heeled footwear (12). Csapo et al., (13) showed using axial-  
83 plane magnetic resonance imaging that experienced wearers were associated with **shortening**  
84 **of the triceps-surae muscle-tendon units, compared to in-experienced users. Leading to the**  
85 **notion that wearing experience may affect female's susceptibility to chronic musculoskeletal**  
86 **pathologies.**

87

88 Previously highlighted analyses concerning the biomechanical effects of high-heeled  
89 footwear on compressive joint loads linked to the aetiology of osteoarthritis, have utilized  
90 joint moments as pseudo indices of global joint kinetics (14). Furthermore, there has yet to be  
91 a comparative examination of muscle tendon unit kinematics when wearing high-heels,  
92 owing to a lack of suitable measurement techniques capable of quantifying muscle  
93 mechanics. Importantly, Herzog et al., (15) showed that muscles are the primary contributors

94 to lower extremity joint loading. Yet the complex role of muscles in controlling joint  
95 biomechanics during human movement has received insufficient attention within the  
96 literature, possibly due to difficulties in calculating muscle kinetics and kinematics. However,  
97 advances in musculoskeletal modelling have led to the development of bespoke software  
98 which allows skeletal muscle force distributions and muscle tendon lengths to be simulated  
99 during movement using motion capture based data (16). To date, such approaches have not  
100 yet been utilized to explore biomechanical differences between high-heeled and traditional  
101 footwear.

102

103 The aims of the current investigation were therefore twofold. Firstly, to examine the effects  
104 of different heeled footwear heights on lower extremity joint loading and triceps-surae  
105 muscle tendon kinematics, using a simulation based approach. Secondly, to examine both  
106 experienced and in-experienced high-heel users, in order to determine whether wearing  
107 experience affects female's potential susceptibility to chronic musculoskeletal pathologies.  
108 The current investigation may provide further important information regarding the potential  
109 chronic effects of high-heeled footwear in experienced and in-experienced users.

110

## 111 **Methods**

### 112 *Participants*

113 Twenty four female participants (12 experienced high-heel wearers; age  $30.54 \pm 5.55$  years,  
114 height  $1.65 \pm 0.08$  cm and body mass  $63.42 \pm 6.73$  kg and 12 inexperienced high-heel wearers;  
115 age  $29.24 \pm 4.78$  years, height  $1.66 \pm 0.11$  cm and body mass  $65.27 \pm 5.98$  kg) volunteered to  
116 take part in this study. To be considered an experienced high-heel wearer, participants had to

117 have worn heels with a minimum heel height of 5 cm at least five times a week for a  
118 minimum of 2 years (11). All participants were free from pathology at the time of data  
119 collection and provided written informed consent, in accordance with the principles outlined  
120 in the Declaration of Helsinki. The procedure utilized for this investigation was approved; by  
121 a university ethical committee (REF 637).

122

### 123 *Experimental footwear*

124 The footwear used during this study consisted of traditional footwear (New Balance 1260 v2;  
125 Figure 1a), high heels (10cm heel; Figure 2d), medium heels (7cm heel; Figure 2c), and low  
126 heels (4cm heel; Figure 2b) in sizes 3–6 in UK. The heeled-footwear were identical with the  
127 exception of the heel heights.

128

129 @@@ *Figure 1 near here* @@@

130

### 131 *Procedure*

132 Participants walked at a velocity of 1.5 m/s ( $\pm 5\%$ ), striking an embedded piezoelectric force  
133 platform (Kistler, Kistler Instruments Ltd) with their right (dominant) foot. Walking velocity  
134 was monitored using infrared timing gates (Newtest, Oy Koulukatu). The stance phase was  
135 delineated as the duration over which 20 N or greater of vertical force was applied to the  
136 force platform. Participants completed a minimum of five successful trials in each footwear  
137 condition. The order that participants walked in each footwear condition was  
138 counterbalanced. Kinematics and ground reaction forces data were synchronously collected.

139 Kinematic data was captured at 250 Hz via an eight camera motion analysis system (Qualisys  
140 Medical AB) and ground reaction forces captured at 1000 Hz. Dynamic calibration of the  
141 motion capture system was performed before each data collection session.

142

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

153

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

159

160 *Data processing*

161 Dynamic trials were digitized using Qualisys Track Manager, in order to identify anatomical  
162 and tracking markers and then exported as C3D files to Visual 3D (C-Motion, Germantown,  
163 MD). All data were normalized to 100 % of the stance phase. Ground reaction force and  
164 kinematic data were smoothed using cut-off frequencies of 12 and 6 Hz with a low-pass  
165 Butterworth 4th order zero lag filter (17). All net force parameters throughout were  
166 normalized by dividing by bodyweight (BW). Following this the external vertical rate of  
167 loading (BW/s) was quantified, as the peak increase in force between adjacent data points.

168

169 Data during the stance phase were exported from Visual 3D into OpenSim 3.3 software  
170 (Simtk.org). A validated musculoskeletal model with 12 segments, 19 degrees of freedom  
171 and 92 musculotendon actuators (18) was used to estimate lower extremity joint forces. The  
172 model was scaled for each participant to account for the anthropometrics of each. As muscle  
173 forces are the main determinant of joint compressive forces (15), muscle kinetics were  
174 quantified using a weighted static optimization in accordance with Steele et al., (19).  
175 Compressive ankle, medial/ lateral tibiofemoral and hip joint forces were calculated via the  
176 joint reaction analyses function using the muscle forces generated from the static  
177 optimization process as inputs. The joint reaction analysis function in OpenSim calculates the  
178 joint loads transferred between two contacting bodies, about the joint centre location  
179 identified during the static trial (19). In the current investigation, hip joint forces were  
180 representative of the sum of contact forces between the femur and acetabular cartilage,  
181 tibiofemoral forces between the medial/ lateral tibial and femoral cartilage and ankle joint  
182 forces between the tibia and talar cartilage. From the above processing, peak ankle force,  
183 peak medial tibiofemoral force, peak lateral tibiofemoral force and peak hip force were  
184 extracted for statistical analyses. In addition ankle, medial/ lateral tibiofemoral and hip

185 instantaneous load rates (BW/s) were also extracted by obtaining the peak increase in force  
186 between adjacent data points.

187

188 Patellofemoral loading was quantified using a model adapted from van Eijden et al., (20). A  
189 key drawback of this model is that co-contraction of the knee flexor musculature is not  
190 accounted for (21). Taking this into account, summed hamstring and gastrocnemius forces  
191 derived from the static optimization procedure were multiplied by their estimated knee joint  
192 muscle moment arms as a function of knee flexion angle (22), and then added together to  
193 determine the knee flexor torque during the stance phase. In addition to this, the knee  
194 extensor torque was also calculated by dividing the summed quadriceps forces by this muscle  
195 groups' knee joint muscle moment arms as a function of knee flexion angle (van Eijden et al.,  
196 (20). The knee flexor and extensor torques were then summed and subsequently divided by  
197 the quadriceps muscle moment arm to obtain quadriceps force adjusted for co-contraction of  
198 the knee flexor musculature. Patellofemoral force was quantified by multiplying the derived  
199 quadriceps force by a constant obtained by using the data of Eijden et al., (20). Finally,  
200 patellofemoral joint stress (KPa/BW) was quantified by dividing the patellofemoral force by  
201 the patellofemoral contact area. Patellofemoral contact areas were obtained by fitting a  
202 polynomial curve to the sex specific data of Besier et al., (23). From the above processing,  
203 peak patellofemoral force and peak patellofemoral stress were extracted for statistical  
204 analyses. In addition, patellofemoral instantaneous load rate (BW/s) was also extracted by  
205 obtaining the peak increase in force between adjacent data points.

206

207 Finally, Achilles tendon forces were estimated in accordance with the protocol of  
208 Almonroeder et al., (24), by summing the muscle forces of the medial gastrocnemius, lateral,

209 gastrocnemius, and soleus muscles. From the above processing, peak Achilles tendon force  
210 and Achilles tendon instantaneous load rate (BW/s) were extracted for statistical analyses.

211

212 Heeled footwear may affect the number of footfalls required to complete a set distance. We  
213 therefore firstly calculated integral of the hip, tibiofemoral, patellofemoral, ankle and  
214 Achilles tendon forces during the stance phase, using a trapezoidal function. In addition to  
215 this, we also estimated the total force per mile (BW·mile) by multiplying these parameters by  
216 the number of steps required to walk one mile. The number of steps required to complete one  
217 mile was quantified using the step length (m), which was determined by taking the difference  
218 in the horizontal position of the foot centre of mass between the right and left legs at  
219 footstrike.

220

221 Muscle–tendon lengths were also determined using OpenSim in accordance with Sinclair,  
222 (25), via the positions of their proximal and distal muscle origins. The muscle tendon  
223 complexes which were evaluated as part of the current research were the Lateral  
224 gastrocnemius, Medial gastrocnemius and Soleus. The mean lengths of these muscle tendon  
225 units during the stance phase were extracted for statistical analysis.

226

### 227 *Statistical analyses*

228 Descriptive statistics of means and standard deviations were obtained for each outcome  
229 measure. Shapiro-Wilk tests were used to screen the data for normality. Differences in  
230 biomechanical parameters were examined using 4 (FOOTWEAR) x 2 (EXPERIENCE)  
231 mixed ANOVA's. In the event of a significant main effect pairwise comparisons were

232 performed. Statistical significance was accepted at the  $P \leq 0.05$  level (26). Effect sizes for all  
233 significant findings were calculated using partial Eta<sup>2</sup> ( $p\eta^2$ ). In accordance with Sinclair et al.,  
234 (26) the minimum clinically important difference (MCID) was considered to be 2.3 \* the  
235 pooled standard error of measurement. All statistical actions were conducted using SPSS  
236 v24.0 (SPSS Inc, Chicago, USA).

237

## 238 **Results**

239 Tables 1-2 and figures 2-3 present the joint load and muscle kinematics variables obtained as  
240 a function of the different heel height conditions and experience in wearing high-heeled  
241 footwear.

242

243 *@@@ Figure 2 near here @@@*

244 *@@@ Figure 3 near here @@@*

245 *@@@ Table 1 near here @@@*

246 *@@@ Table 2 near here @@@*

247

### 248 *Spatiotemporal and loading rate parameters*

249 A main effect of FOOTWEAR was found for step length ( $P < 0.05$ ,  $p\eta^2 = 0.51$ ). Post-hoc  
250 pairwise comparisons showed that step length was significantly greater in the trainer,  
251 compared to the high, medium and low heels, and significantly larger in the medium and low  
252 heels compared to the high heel condition (Table 1).

253

254 A main effect of FOOTWEAR was found for the external vertical load rate ( $P<0.05$ ,  
255  $p\eta^2=0.50$ ). Post-hoc pairwise comparisons showed that the load rate was significantly greater  
256 in the high heels compared to the, medium, low and trainer conditions, and significantly  
257 larger in the medium and low heels compared to the trainer (Table 1).

258

#### 259 *Hip joint loading*

260 For the load experienced per mile, a main effect of FOOTWEAR ( $P<0.05$ ,  $p\eta^2=0.19$ ) was  
261 observed. Post-hoc pairwise comparisons showed that the load experienced per mile was  
262 significantly larger in the high and low heels in comparison to the trainer (Table 1).

263

#### 264 *Tibiofemoral joint loading*

265 For medial tibiofemoral load rate, a main effect of FOOTWEAR ( $P<0.05$ ,  $p\eta^2=0.15$ ) was  
266 observed. Post-hoc pairwise comparisons showed that the medial tibiofemoral load rate was  
267 significantly larger in the high heels in comparison to the low heels and trainer conditions  
268 (Table 1). For medial tibiofemoral load experienced per mile, a main effect of FOOTWEAR  
269 ( $P<0.05$ ,  $p\eta^2=0.23$ ) was observed. Post-hoc pairwise comparisons showed that the load  
270 experienced per mile was significantly larger in the high, medium and low heel conditions in  
271 comparison to the trainer (Table 1).

272

273 For lateral tibiofemoral load experienced per mile, a main effect of FOOTWEAR ( $P<0.05$ ,  
274  $p\eta^2=0.24$ ) was observed. Post-hoc pairwise comparisons showed that the load experienced

275 per mile was significantly larger in the high heel in comparison to the medium, low and  
276 trainer conditions (Table 1).

277

### 278 *Patellofemoral joint loading*

279 Main effects of FOOTWEAR were observed for peak patellofemoral force ( $P<0.05$ ,  
280  $p\eta^2=0.70$ ) and stress ( $P<0.05$ ,  $p\eta^2=0.68$ ). Post-hoc pairwise comparisons showed that peak  
281 force and stress were significantly greater in the high, medium and low heels in comparison  
282 to the trainer, and significantly larger in the high heels compared to the medium and low heel  
283 conditions (Table 1; Figure 2de). In addition, a main effect of FOOTWEAR was observed for  
284 patellofemoral load rate ( $P<0.05$ ,  $p\eta^2=0.61$ ). Post-hoc pairwise comparisons showed that load  
285 rate was significantly greater in the high, medium and low heels in comparison to the trainer,  
286 and significantly larger in the high heels compared to the medium and low heel conditions  
287 (Table 1). Finally, a significant main effect of FOOTWEAR was observed for patellofemoral  
288 force per mile ( $P<0.05$ ,  $p\eta^2=0.63$ ). Post-hoc pairwise comparisons showed that each footwear  
289 differed significantly from one another, with the patellofemoral force per mile increasing  
290 linearly with increases in heel height (Table 1).

291

### 292 *Ankle joint loading*

293 Main effects of FOOTWEAR were observed for peak ankle force ( $P<0.05$ ,  $p\eta^2=0.60$ ) and  
294 Achilles tendon force ( $P<0.05$ ,  $p\eta^2=0.82$ ). Post-hoc pairwise comparisons showed that each  
295 footwear differed significantly from one another, with peak ankle force decreasing linearly  
296 with increases in heel height (Table 1; Figure 2fg).

297

298 In addition, a significant main effect of FOOTWEAR was observed for ankle force per mile  
299 ( $P < 0.05$ ,  $\eta^2 = 0.46$ ). Post-hoc pairwise comparisons showed that ankle force per mile was  
300 significantly greater in the medium, low and trainer conditions compared to the high heels.  
301 Furthermore, it was also revealed that force per mile was significantly larger in the low heel  
302 and trainer compared to the medium condition (Table 1). Finally, a significant main of  
303 FOOTWEAR was observed for Achilles tendon force per mile ( $P < 0.05$ ,  $\eta^2 = 0.73$ ). Post-hoc  
304 pairwise comparisons showed that each footwear differed significantly from one another,  
305 with Achilles tendon force per mile decreasing linearly with increases in heel height (Table  
306 1).

307

### 308 *Muscle lengths*

309 There were FOOTWEAR main effects for the Soleus ( $P < 0.05$ ,  $\eta^2 = 0.80$ ), Medial  
310 gastrocnemius ( $P < 0.05$ ,  $\eta^2 = 0.85$ ) and Lateral gastrocnemius ( $P < 0.05$ ,  $\eta^2 = 0.85$ ). Post-hoc  
311 pairwise comparisons showed for each muscle, that each footwear differed significantly from  
312 one another, with the mean muscle lengths decreasing linearly with increases in heel height  
313 (Table 2; Figure 3a-c).

314

### 315 **Discussion**

316 The current study examines the effects of different high-heeled footwear heights on lower  
317 extremity joint loading and triceps-surae muscle tendon kinematics. To the authors  
318 knowledge this represents the first investigation to examine the biomechanics of high-heeled  
319 footwear using musculoskeletal simulation, and may provide more detailed information  
320 regarding the effects of high-heeled footwear in experienced and in-experienced users.

321

322 The current investigation showed that compressive hip joint loading experienced per mile  
323 was significantly increased in the high and low heels in comparison the trainer. As no  
324 alterations in peak loading were observed in these conditions, it can be concluded that the  
325 increased loading was mediated as a function of the decreased step length. The initiation and  
326 progression of osteoarthritis is mediated through chronic compressive loading experienced at  
327 the joint itself (28). However, whilst the current investigation showed that there were  
328 statistical increases in compressive hip loading, the magnitude of the differences between  
329 footwear conditions did not exceed the MCID. This leads to the conclusion that heeled  
330 footwear may not influence wearers' susceptibility to chronic hip joint pathology, although  
331 further analysis should seek to confirm this notion.

332

333 In addition, the current investigation showed that compressive joint loading at both the  
334 medial and lateral aspects of the tibiofemoral joint were statistically influenced as a function  
335 of the different experimental footwear. At the medial tibiofemoral compartment, the load rate  
336 was larger in the high heels compared to the low heels and trainer conditions, and the load per  
337 mile was greater in each of the heeled footwear in compared to the trainer. However, only  
338 differences in force per mile between the high heels and trainer were beyond the MCID  
339 threshold. At the lateral compartment, the loads experienced per mile were greater in the high  
340 heel compared to the medium, low and trainer conditions, although the magnitude of these  
341 differences did not exceeded the MCID. Once again as no alterations in peak loading were  
342 observed, it can be concluded that increased loads per mile were mediated as a function of  
343 decreases in step length. However, the increased medial load rate in the heeled footwear is  
344 likely a consequence of the increased rate at which the external ground reaction force is

345 experienced in these conditions. The medial tibiofemoral compartment is at much greater risk  
346 from degenerative joint osteoarthritis compared to the lateral aspect of the knee joint (29). As  
347 such it appears that irrespective of experience, the high heels increased the risk of medial  
348 knee osteoarthritis compared to the trainer, an observation in agreement with the propositions  
349 of Kerrigan et al., (10).

350

351 At the patellofemoral joint, compressive loading parameters were shown to generally increase  
352 linearly with increases in heel height, and predominantly exceeded the MCID between each  
353 footwear condition. Patellofemoral pain is one of the most common chronic musculoskeletal  
354 disorders of the lower extremities, and like osteoarthritis is more common in females  
355 compared to males (30). Patellofemoral pain may be the result of increased patellofemoral  
356 joint stress (31), and is thought longitudinally, to progress to patellofemoral joint  
357 osteoarthritis (32). The enhanced patellofemoral joint stress shown in the high-heeled  
358 footwear conditions was mediated by increases in the patellofemoral force, in particular as  
359 increases in knee flexion shown in the high-heeled footwear conditions (Supplemental data  
360 1a) lead to increased patellofemoral contact areas (23). In turn it is proposed that the  
361 augmented patellofemoral force was caused by increases in knee extensor muscle forces, a  
362 key input parameter into the patellofemoral joint musculoskeletal model. Increased knee  
363 extensor muscle force requirements were mediated via a more posterior orientation of the  
364 ground reaction force vector in the high-heeled footwear (33). The current study therefore  
365 provides strong evidence that high-heeled footwear of increased height results in elevated  
366 patellofemoral joint stress, which could potentially lead to an increase in patellofemoral  
367 symptoms over time.

368

369 It was also revealed that muscle tendon kinematics were significantly influenced as a function  
370 of different heel heights, and importantly that the magnitude of the differences exceeded the  
371 MCID in all cases. Specifically, the current study showed that each of the triceps-surae  
372 muscle tendon-unit lengths during the stance phase decreased linearly with increases in heel  
373 height. This investigation also showed that ankle and Achilles tendon loading also decreased  
374 linearly alongside increases in heel height, with the differences between footwear conditions  
375 surpassing the magnitude of the MCID in the majority of cases. This observation opposes  
376 previous suggestions (5, 6), who suggested that triceps-surae muscles forces are likely to  
377 increase when wearing high-heels. It is proposed that the decreased ankle and Achilles tendon  
378 loading can be explained concomitantly by the shorter triceps-surae muscle lengths, reduced  
379 Achilles tendon moment arm as a function of enhanced ankle plantar flexion, combined with  
380 a ground reaction force vector that passes closer to the ankle joint centre. These parameters  
381 serve to increase the forces generated by the triceps-surae muscles in the trainer condition  
382 (Supplemental data 1bcd), which strongly govern the loads experienced compressively by  
383 ankle joint and are solely responsible for those experienced by the Achilles tendon. This  
384 finding does oppose the notion proposed by Csapo et al., (13) that increased Achilles tendon  
385 cross-sectional area revealed in experienced high-heel wearers is mediated via increases in  
386 the relative muscle forces acting on the tendon-aponeurosis complex. Future analyses should  
387 therefore seek to better understand examine the biomechanical mechanisms that promote  
388 Achilles tendon hypertrophy in regular high-heel wearers.

389

390 However, the linear reductions in muscle tendon lengths strongly support the findings of  
391 Csapo et al., (13), who showed that regular usage of high-heeled footwear placed the triceps-  
392 surae muscles in a chronically shortened position, and are attributable to the ankle being at an  
393 increasingly more plantarflexed angle. Acute shortening of the triceps-surae muscle tendon

394 units during walking is energetically inefficient (13), as it causes unnecessary overlap of the  
395 actin-myosin units and forces the muscle fibers into a non-optimal operating range (34).  
396 Habitual shortening of the triceps-surae muscle tendon units through regular utilization of  
397 heeled-footwear mediates chronic muscle tendon unit adaptations, whereby the muscle itself  
398 is shortened by reducing the number of in-series sarcomeres in order to transfer the actin-  
399 myosin overlap back to optimal operating range (34).

400

401 Importantly, the current investigation also revealed that there were no statistical main effects  
402 for EXPERIENCE, nor were there any significant interactions between FOOTWEAR x  
403 EXPERIENCE. This observation concurs with those of Ebbeling et al., (34) and Simonsen et  
404 al., (6) yet opposes the observations of Barton et al., (36); de Oliveira Pezzan et al., (37); and  
405 Gefen et al. (38). Nonetheless, the current investigation has shown that heeled footwear is  
406 associated with increased compressive tibiofemoral and patellofemoral joint loading and also  
407 places each of the triceps-surae muscle tendon units in a shortened position during the stance  
408 phase. These parameters are linked to the aetiology of degenerative joint osteoarthritis (28),  
409 and chronic shortening of the triceps-surae muscle tendon units (13). As such the current  
410 investigation indicates that the potential chronic effects of heeled footwear of increasing  
411 heights, appear to be independent of the users experience in wearing high-heeled footwear.

412

413 In conclusion, although walking biomechanics in heeled-footwear has received previous  
414 research attention; there has yet to be a quantitative comparison of lower extremity joint  
415 loading/ muscle tendon kinematics, using a musculoskeletal simulation based approach. The  
416 present investigation adds to the current knowledge, by examining the effects of different  
417 high-heeled footwear heights on lower extremity joint loading and triceps-surae muscle

418 tendon kinematics in experienced and in-experienced heel wearers. This investigation showed  
419 irrespective of experience, that compressive loading at the medial tibiofemoral and  
420 patellofemoral joints was enhanced beyond the MCID in high-heels of increasing height.  
421 Furthermore, irrespective of experience, the triceps-surae muscle tendon units were shown to  
422 be placed in a shortened position when wearing high-heels of increasing height, with the  
423 magnitude of the differences exceeding the MCID. It can therefore be concluded that heeled-  
424 footwear increase the mechanical factors linked to the aetiology of degenerative joint  
425 osteoarthritis and chronic shortening of the triceps-surae muscle tendon units. Therefore, the  
426 current investigation provides evidence that irrespective of experience, heeled-footwear of  
427 increasing height may negatively influence females' lower extremity musculoskeletal health.

428

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535

536 **List of figures**

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547 = medium heel, black dot = low heel, dark grey = trainer, black dash = high heel experienced,  
548 black outline = medium heel experienced, grey dot = low heel experienced and dark grey  
549 outline = trainer experienced).

550

### 551 **Supplemental data**

552 Appendix Figure 1: (a. = knee flexion angle during the stance phase, b. = Soleus muscle  
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554 (black = high heel, light grey = medium heel, black dot = low heel, dark grey = trainer, black  
555 dash = high heel experienced, black outline = medium heel experienced, grey dot = low heel  
556 experienced and dark grey outline = trainer experienced)