

1 **The influence of foot-strike technique on the neuromechanical function of the foot**

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3 Dr Luke A Kelly<sup>1</sup>, Dr Dominic J Farris<sup>1</sup>, A/Professor Glen A Lichtwark<sup>1</sup> & Professor Andrew

4 G Cresswell<sup>1</sup>

5 *1 - School of Human Movement and Nutrition Sciences, The University of Queensland,*

6 *Australia*

7

8 Corresponding Author:

9 Dr Luke Kelly

10 School of Human Movement and Nutrition Sciences

11 The University of Queensland

12 Blair Drive

13 St Lucia

14 QLD, Australia

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27 **Abstract**

28 *Purpose:* The aim of this study was to investigate the influence of foot-strike technique on  
29 longitudinal arch mechanics and intrinsic foot muscle function during running.

30 *Methods:* 13 healthy participants ran barefoot on a force-instrumented treadmill at  $2.8\text{ms}^{-1}$  with  
31 a forefoot (FFS) and rear-foot (RFS, habitual) running technique, while kinetic, kinematic and  
32 electromyographic (EMG) data from the intrinsic foot muscles were collected simultaneously.  
33 The longitudinal arch was modeled as a single “mid-foot” joint representing motion of the rear-  
34 foot (calcaneus) relative to the forefoot (metatarsals). An inverse dynamic analysis was  
35 performed to estimate joint moments generated about the mid-foot, as well as mechanical work  
36 and power.

37 *Results:* The mid-foot was more plantar flexed (higher arch) at foot contact when running with  
38 a forefoot running technique (RFS  $0.2 \pm 1.8^\circ$  v FFS  $6.9 \pm 3.0^\circ$ , ES = 2.7), however there was no  
39 difference in peak mid-foot dorsiflexion in stance (RFS  $-11.6 \pm 3.0^\circ$  v FFS  $-11.4 \pm 3.4^\circ$ , ES =  
40 0.63). When running with a forefoot technique, participants generated greater moments about  
41 the mid-foot (27% increase, ES = 1.1) and performed more negative work (240% increase, ES  
42 = 2.2) and positive work (42% increase, ES = 1.1) about the mid-foot. Stance phase muscle  
43 activation was greater for Flexor Digitorum Brevis (20% increase, ES = 0.56) and Abductor  
44 Hallucis (17% increase, ES = 0.63) when running with a forefoot technique.

45 *Conclusion:* Forefoot running increases loading about the longitudinal arch and also increases the  
46 mechanical work performed by the intrinsic foot muscles. These findings have substantial  
47 implications in terms of injury prevention and management for runners who transition from a rear-  
48 foot to a forefoot running technique.

49 **Keywords**

50 Intrinsic foot muscles, longitudinal arch, forefoot running, plantar fascia, foot biomechanics

51

## 52 **Introduction**

53 The orientation of a runner's foot at the moment of foot-ground impact is often employed as a  
54 method to classify running technique (2, 7, 29, 33). Foot-strike patterns are generally classified  
55 in three groups; (1) runners who land in a heel-first manner (rear-foot strike), (2) runners who  
56 land on the forefoot (forefoot strike), and (3) runners who land with a simultaneous heel and  
57 forefoot contact (mid-foot strike) (3, 7). Although the majority (>75%) of distance runners  
58 adopt a rear-foot running technique (14, 25), it has been suggested that a forefoot running  
59 technique is mechanically advantageous, as it affords the runner greater opportunity to recycle  
60 the energy associated with foot-ground impact via elastic stretch and recoil of the tendons and  
61 ligaments of the ankle and foot (28, 29, 33). Consequently, the popularity of running "re-  
62 training" programs that promote a forefoot landing pattern to reduce injury risk and improve  
63 running performance have surged (3, 10, 15).

64

65 The longitudinal arch (LA) of the human foot behaves in a spring-like manner when running,  
66 temporarily storing and then subsequently returning a considerable portion of the mechanical  
67 energy required for each stride (22, 36, 39). Compression (lowering and lengthening) of the LA  
68 during the first half of stance phase allows mechanical energy to be temporarily stored within  
69 the stretched ligaments, muscles and tendons that span this structure (20, 22, 39). This energy  
70 is subsequently returned in late stance, as the resultant ground reaction force (GRF) declines  
71 and the stretched elastic structures shorten to allow the LA to recoil (rise and shorten) (20, 22,  
72 33).

73

74 Recently studies have explored the hypothesis that a forefoot running technique enhances foot-  
75 spring function by allowing a greater proportion of mechanical energy to be recycled during

76 each stride (30, 33, 39). Consistent with the hypothesis of enhanced foot-spring function, these  
77 studies have all reported an increase in LA compliance with a forefoot technique, compared to  
78 a rear-foot technique, when running at matched velocities. However, the increased compliance  
79 appears to be due to a higher LA at foot contact, rather than an increase in peak LA compression  
80 in mid-stance (30, 33, 39). A key finding in each of these studies was that peak LA compression  
81 and model derived plantar aponeurosis strains were similar, regardless of foot-strike technique  
82 (30, 33, 39). Given that peak strain primarily determines the magnitude of stored elastic energy,  
83 this finding suggests no energetic benefits to a forefoot running technique. The studies by  
84 McDonald et al. (2016) and Wager & Challis (2016) provide valuable insight into the  
85 biomechanics of the LA when running with rear-foot and forefoot running techniques.  
86 However, these studies have a methodological limitation in their assumption that the foot is a  
87 passive structure with no active muscular control of its biomechanical function.

88  
89 Forefoot running is characterized by the centre of pressure (COP) being located further anterior  
90 in the foot and higher peak GRF (3, 6, 7, 31). Thus in a similar manner to the ankle joint (35),  
91 an increase in the length of the GRF moment arm, combined with the higher peak GRF, will  
92 likely increase loading of the LA when running with a forefoot technique. Given that the  
93 intrinsic foot muscles have similar anatomical pathways to the plantar aponeurosis (24, 27, 38)  
94 and have the capacity to alter the stiffness of the LA (18, 20) it is highly likely that these muscles  
95 also contribute to the observed alterations in LA mechanics between foot-strike techniques. An  
96 increase in activation of the intrinsic foot muscles when running with a forefoot technique could  
97 potentially reduce excessive LA compression and reduce plantar aponeurosis strain, providing  
98 an explanation for the lack of difference in plantar aponeurosis strain observed in previous  
99 studies (30, 39). Furthermore, these muscles are also known to display preparatory activation  
100 prior to foot contact (20, 21), thus it is conceivable that increased activation of these muscles in

101 late swing may also explain the increased LA height at foot-contact when running with a  
102 forefoot technique.

103

104 It is currently unknown if the intrinsic foot muscles contribute to the regulation of LA motion  
105 when switching from a rear-foot to a forefoot running technique. It is also unknown if loading  
106 of the LA changes when converting to forefoot running technique. Therefore, the aim of this  
107 study was to test the hypothesis that a forefoot running technique results in higher magnitudes  
108 of activation in the intrinsic foot muscles during the stance and swing phases while running,  
109 due to the higher mechanical loading occurring about the LA.

110

## 111 **Methods**

### 112 Participants

113 Thirteen healthy participants (five females and eight males, age  $24 \pm 6$  years; height:  $172 \pm 8$   
114 cm; mass:  $72 \pm 12$  kg) with no diagnosed history of lower limb injury in the previous six months  
115 or known neurological impairment volunteered to participate in the study. All participants were  
116 habitually shod recreational runners, who classified themselves as non-forefoot strikers.  
117 Written informed consent was obtained from each subject. The study protocol was approved  
118 by the institutional human research ethics committee and conducted in accordance with the  
119 Declaration of Helsinki.

120

### 121 Experimental Protocol

122 Following a 3-min warm up period and familiarization procedure, participants ran barefoot on  
123 a force-instrumented treadmill (AMTI, force-sensing tandem treadmill, Watertown, MA, USA)  
124 at  $2.8 \text{ m}\cdot\text{s}^{-1}$  with their habitual foot-strike (HFS) pattern and a forefoot strike pattern (FFS), in  
125 a counter-balanced order. During the familiarization session, foot-strike pattern was confirmed

126 by the investigator, using visual analysis. None of the participants displayed a habitual forefoot  
127 running technique. Prior to commencement of each experimental condition, participants were  
128 given as much time as necessary to familiarize themselves with running on the treadmill with  
129 a forefoot strike pattern. Participants were considered familiarized with the forefoot running  
130 technique once they were able to perform 20 seconds of consecutive forefoot strikes at the  
131 experimental speed and reported to be comfortable running with this technique. Foot-strike  
132 pattern during the familiarisation period was confirmed visually by the investigator (LK).  
133 Kinetic, kinematic and electromyographic (EMG) data were collected simultaneously for  
134 approximately 15-20 strides (toe-off to ipsilateral toe-off) across a 15 s data collection period  
135 for each running condition (HFS and FFS).

136

## 137 Data Acquisition

### 138 *Kinematic and kinetic measurements*

139 Motion data was captured at 200 Hz using an eight-camera 3D optoelectronic motion capture  
140 system (Oqus, Qualysis, Gothenburg, Sweden). GRF and EMG data were synchronously  
141 captured with the motion data at 4000 Hz using a 14-bit analogue to digital converter using and  
142 the Qualisys Track Management software (Qualisys, Gothenburg, Sweden).

143

144 Three-dimensional (3D) motion of the foot and shank was modelled from motion capture data  
145 during each running trial. In order to quantify motions of the foot segment and the LA (21, 26)  
146 retro-reflective markers (9.0 mm diameter) were secured on the skin of the right foot overlying  
147 the medial and lateral malleoli, posterior calcaneus, navicular tuberosity and head of the first  
148 and fifth metatarsals. Additional markers were applied to the medial and lateral femoral  
149 condyles and a rigid cluster of four markers was placed on the antero-lateral aspect of the shank.  
150 During a standing calibration trial, markers located on the segment endpoints were used to

151 generate a two-segment model of the shank and foot. A virtual marker (med-knee)  
152 corresponding to the position of the medial marker was also generated in order to reduce the  
153 influence of soft tissue artefact on skin-mounted markers located at the knee. Following the  
154 calibration trial, the medial and lateral knee markers were removed and the motion of the shank  
155 and med-knee virtual marker was tracked using the rigid marker cluster located on the shank.  
156 Markers were adhered with double sided adhesive and further secured with cohesive bandage,  
157 allowing secure positioning for both running conditions. Markers were not removed between  
158 experimental trials.

159

### 160 *Electromyography*

161 Identification of the abductor hallucis (AH) and flexor digitorum brevis (FDB) muscles in the  
162 right foot of each subject was conducted using real-time B-mode ultrasound imaging (10 MHz  
163 linear array, Ultrasonix RP, USA). Subsequently, bi-polar fine-wire electrodes (0.051 mm  
164 stainless steel, Teflon coated wires, Chalgren, USA) with a detection length of 4 mm and inter-  
165 electrode distance of 4 mm were inserted using delivery needles (0.5 mm x 50 mm) into the  
166 muscle tissue of AH and FDB under ultrasound guidance, in accordance with previously  
167 described methods (19). Sterile techniques were used for the insertion of all wires. Surface  
168 EMG data was additionally collected from medial gastrocnemius (MG) and soleus (SOL) from  
169 the right leg of all participants using Ag-AgCl electrodes with a diameter of 10 mm and an  
170 inter-electrode distance of 20 mm (Tyco Healthcare Group, Neustadt, Germany). A surface  
171 reference electrode, of the same type and size, was placed over the right fibula head. Prior to  
172 electrode placement, the areas of the leg corresponding to the electrode placement sites were  
173 shaved, lightly abraded and cleaned with isopropyl alcohol. All EMG signals were amplified  
174 1000 times and hardware filtered with a bandwidth of 30 to 1000 Hz (MA300, Motion Labs,  
175 LA, USA). In order to minimize movement artefacts, the fine-wire electrodes, surface

176 electrodes, connectors, cabling and pre-amplifiers were secured with cohesive bandage around  
177 the foot and shank.

178

179 Prior to data collection, each participant was asked to perform foot manoeuvres known to  
180 activate each of the muscles separately. Activation of the AH muscle was assessed by asking  
181 the participant to abduct and flex the great toe about the metatarso-phalangeal (MTP) joint,  
182 while avoiding flexion of the interphalangeal (IP) joint. The FDB activation was assessed by  
183 asking the participant to flex toes 2-4 about the MTP joints, while minimising flexion of the  
184 distal IP joints (Kelly, Kuitunen, Racinais, & Cresswell, 2012; Kelly, Racinais, & Cresswell,  
185 2013). When predicted EMG patterns could be detected, it was concluded that the fine-wire  
186 electrodes were in the correct location. If not, the electrodes were withdrawn by approximately  
187 1 mm until appropriate activation patterns could be detected and possible crosstalk excluded.  
188 In order to confirm signal integrity and quality between trials, the EMG signals were re-tested  
189 following each trial using the same foot manoeuvres applied above. A Velcro strap was secured  
190 around the participant's waist, which enabled the EMG amplifier box to be secured to the  
191 subject without interfering with their gait. A lightweight optical cable connected the amplifier  
192 box to the analogue to digital converter that was positioned well away from the subject and  
193 treadmill.

194

195 Data analysis

196 Marker trajectories and GRF data were exported to Matlab (Matlab, The Mathworks Inc., USA)  
197 while EMG data files were exported to Visual3D (C-motion Inc., Germantown, MD, USA) for  
198 post processing and analysis.

199

200 *Ground reaction forces*



201 GRF data was filtered using a 35 Hz recursive, low-pass second order Butterworth filter. A  
202 vertical GRF threshold was set to define each toe-off as occurring when vertical GRF fell below  
203 50 N, while foot contact was defined as occurring when vertical force rose above 50 N. The  
204 magnitude of the peak vertical and antero-posterior (A-P) components of the GRF were  
205 calculated and normalized to body weight (BW) for each participant. Peak propulsive force was  
206 defined as the peak positive value of the A-P component of the GRF.

207

#### 208 *Stride temporal characteristics*

209 Swing phase was defined as the period from right toe-off to right foot contact, while stance  
210 phase was defined as occurring between right foot contact and right toe-off. One stride cycle  
211 was considered as right toe-off to the subsequent right toe-off based on GRF data. Stride  
212 duration and ground contact times were calculated for both conditions, for all participants.

213

#### 214 *Joint kinematics*

215 Prior to calculation of joint kinematics, marker trajectories were low pass filtered using a  
216 recursive second order Butterworth filter at 20 Hz. Initially, an assumed rigid segment was  
217 created for the foot (21, 26). All marker data were transformed to the foot segment co-ordinate  
218 system, allowing all sagittal plane angles to be represented relative to the orientation of the z-  
219 axis of the foot. Thus, all sagittal plane rotations represent flexion-extension of their respective  
220 joints. Ankle angle was defined as the planar angle created by the bisection of a vector  
221 projecting from the medial malleolus to the med-knee virtual marker, with another vector  
222 projecting from the head of the first metatarsal to the medial malleolus, projected onto the  
223 sagittal plane of the foot. Ankle angle was offset to the ankle angle during quiet bipedal stance,  
224 calculated during the standing calibration trial. Foot-strike angle was defined as the sagittal  
225 plane ankle angle at foot contact. Minimum ankle angle was defined as the minimum angle

226 during stance phase, representing peak ankle dorsiflexion. Ankle angle at toe-off was also  
227 calculated.

228

229 In order to gain a deeper understanding of the mechanical behaviour of the foot, we also  
230 modelled the foot as a two-segment structure with rear-foot and mid-foot joints. The rear-foot  
231 joint angle was defined as the internal angle created by the bisection of a vector projecting from  
232 the med-knee virtual marker to the medial malleolus with another vector projecting from the  
233 navicular marker to the medial malleolus marker, projected onto the sagittal plane of the foot  
234 (Figure 1). Positive rear-foot joint angles indicate plantar-flexion of the rear-foot relative to the  
235 shank. Thus the rear-foot angle is representative of the ankle joint in our multi-segment foot  
236 model.

237

238 The LA is a complex structure comprised of a series of articulations made by a number of  
239 different bones (16, 17). Given the difficulties in quantifying the rotations about a series of  
240 small joints, located in such close proximity, the LA is often modelled as a single functional  
241 ‘mid-foot’ joint, representing the generalised angular motion of this structure (5, 11, 20, 21, 26,  
242 39). Therefore the biomechanical function of the LA during running was represented by a ‘mid-  
243 foot’ joint, defined as the internal angle created by the bisection of a vector projecting from the  
244 medial malleolus marker to the navicular marker (rear-foot segment) and another vector  
245 projecting from the head of the first metatarsal to the navicular marker, projected onto the  
246 sagittal plane of the foot (26) (forefoot segment, Figure 1). A decrease in mid-foot angle  
247 signifies dorsiflexion, or LA compression (upwards rotation of the distal end of the forefoot  
248 segment) relative to the rear-foot. The rear-foot and mid-foot angles were offset to the  
249 corresponding angles calculated during quiet bipedal stance, so that 0° represent the angle in

250 quiet bipedal stance. Angle at foot contact, minimum stance angle (peak dorsiflexion) and angle  
251 at toe-off were calculated separately for the rear-foot and mid-foot joints, for all participants.

252

### 253 *Joint Kinetics*

254 An inverse dynamic analysis was performed using custom written software (Matlab, The  
255 Mathworks Inc., USA), allowing estimation of net joint moments about the ankle, rear-foot and  
256 mid-foot during stance phase. Unfiltered marker trajectories, ground reaction force and centre  
257 of pressure data were expressed relative to the foot co-ordinate system. Subsequently, proximal  
258 segment forces were resolved from the ground up using a link segment model (40). The mid-  
259 foot segment was defined as the vector projecting from the navicular marker to the marker  
260 located on the head of the first metatarsal, projected onto the sagittal plane of the foot. The rear-  
261 foot segment was defined as the vector projecting from the medial malleolus marker to the  
262 navicular marker, projected onto the sagittal plane of the foot. A traditional rigid foot segment  
263 was also defined in order to estimate ankle joint moments, enabling comparison of multi-  
264 segment foot kinetics with traditional calculations based on the assumption of a rigid foot  
265 segment. Segment masses and inertial properties were incorporated from previous literature (9,  
266 11), with the centre of mass location for each segment being defined as 50% of the distance  
267 between the segment end points (11). Each joint was modelled as a pin joint with rotation  
268 occurring about the joint created by the bisection of the two vectors (Figure 1). Each joint axis  
269 was oriented parallel to the z-axis of the foot co-ordinate system. Following estimation of joint  
270 moments, joint powers were calculated for the stance phase by multiplying the net joint moment  
271 by the angular velocity of each corresponding joint (40). During early stance in a rear-foot  
272 running technique, GRF is applied to the rear-foot and forefoot segments simultaneously when  
273 the foot is flat on the ground. In order to overcome the difficulty in determining the relative  
274 percentage of GRF that is acting on the forefoot and rear-foot segments during early stance,

275 joint moments for the rear-foot and mid-foot were only calculated during stance phase after the  
276 COP had progressed distal to the axis of the mid-foot joint, minimising any potential errors  
277 associated with segment force assignments (37). Subsequently, joint moments and joint powers  
278 for the ankle, rear-foot and mid-foot were filtered using a 20 Hz recursive, second order low-  
279 pass Butterworth filter (Edwards, Troy, & Derrick, 2011). Net negative and positive work  
280 performed by each joint during stance phase was calculated by applying a trapezoidal  
281 integration of joint power curves across the stance phase. Mid-foot stiffness was quantified by  
282 dividing the change in mid-foot joint moment by the change in mid-foot angle during stance phase  
283 (Farley & Morgenroth, 1999).

284

#### 285 *Muscle activation*

286 All EMG signals were high-pass filtered using a recursive second order Butterworth filter at 35  
287 Hz to remove any unwanted low-frequency movement artefact. Subsequently, EMG signals  
288 were visually inspected in order to identify any remaining artefact, which was defined as an  
289 abnormal spike in the signal, typically associated with foot contact. Any remaining artefacts  
290 resulted in the EMG data for that particular stride being excluded from further analysis.  
291 Following DC-offset removal, root mean square (RMS) signal amplitude was calculated across  
292 all EMG data using a moving window of 50 ms to generate an EMG envelope. The EMG  
293 envelope for each muscle was then normalized to the maximum RMS amplitude found across  
294 all conditions for the respective muscle. The normalized peak EMG RMS envelope amplitude  
295 was calculated during stance phase, additionally the normalized mean EMG RMS was  
296 calculated during both the stance and swing phases, allowing comparisons in magnitude of both  
297 stance and swing phase muscle activation between HFS and FFS conditions.

298

299

300 *Statistics*

301 For each individual, the motion, force and EMG data were time normalized and ensemble  
302 averaged across a minimum of 8 stride cycles to form individual variable means for each  
303 condition. Paired Student's t-tests were used to describe the influence of foot-strike pattern on  
304 stride temporal characteristics, peak ground reaction forces, joint kinematics and kinetics, as  
305 well as muscle activation. Statistical differences were established at  $P \leq 0.05$ . Effect sizes (ES)  
306 are presented as standardized mean differences (Cohen's D) with  $\leq 0.70$  representing a small  
307 effect,  $>0.70$  and  $<1.2$  representing a moderate effect, and  $\geq 1.2$  representing a large effect  
308 (Hopkins, 2017). Results are presented as mean  $\pm$  standard deviation (SD) unless otherwise  
309 stated.

310

311 **Results**

312 *Stride temporal characteristics*

313 Participants ran with similar stride durations (ES = 0.07,  $P = 0.54$ ) and ground contact times  
314 (ES = 0,  $P = 0.77$ , Table 1), regardless of the orientation of their ankles at ground contact.

315

316 *Ground reaction forces*

317 When running with a habitual foot strike pattern, vertical GRF- time profiles displayed a  
318 characteristic initial early impact transient, followed by a second, much later, larger peak  
319 (Figure 2). The initial impact peak transient seen for habitual foot strike running was absent  
320 when the participants ran with a forefoot technique (Figure 2), however the magnitude of the  
321 peak vertical GRF was significantly higher with the forefoot technique (ES = 0.81,  $P \leq 0.05$ ,  
322 Table 1). Propulsive and breaking force was similar between conditions (Figure 2).

323

324 *Joint kinematics*

325 When running with a habitual foot strike pattern, participants generally made initial ground  
326 contact with the heel or mid-foot. One participant landed with the ankle in slight plantar  
327 flexion (5.4°), while the remaining participants landed with ankle the ankle in slight  
328 dorsiflexion (range -8.5° to -2.0°, Figure 3). None of the participants landed with a forefoot  
329 strike as their habitual foot-strike pattern. Participants landed with their ankles in a  
330 significantly more plantar flexed orientation when instructed to run with a forefoot strike  
331 (range 6.4° to 20.6°, ES = 3.5,  $P \leq 0.05$ , Table 2). Peak ankle dorsiflexion occurred in mid-  
332 stance for both conditions and was significantly greater when running with a habitual foot-  
333 strike (ES = 0.88,  $P \leq 0.05$ , Table 2). At toe-off the ankle was in a more plantar flexed  
334 orientation in the forefoot running condition (ES = 1.1,  $P \leq 0.05$ , Table 2).

335

336 The rear-foot rotated in a similar manner to the ankle, with dorsiflexion occurring from  
337 forefoot contact through to mid-stance and plantar flexion occurring from mid-stance to toe-  
338 off. The rear-foot was in a more plantar flexed orientation at foot contact when running with a  
339 forefoot strike (ES = 1.1,  $P \leq 0.05$ , Table 2 and Figure 3). Peak rear-foot dorsiflexion  
340 occurred in mid-stance and was reduced in magnitude when running with a forefoot strike  
341 pattern (ES = 3.5,  $P \leq 0.05$ , Table 2). The rear-foot was more plantar flexed at toe-off with the  
342 forefoot running technique (ES = 1.1,  $P \leq 0.05$ , Table 2).

343

344 For both running conditions, the mid-foot dorsiflexed (LA compression) from early stance  
345 through to mid-stance (Figure 3) and subsequently plantar flexed through to toe-off. At foot  
346 contact, the mid-foot joint was more plantar flexed when running with a forefoot strike (ES =  
347 2.7,  $P \leq 0.05$ , Table 2), indicating a higher LA at foot contact. However, there was no  
348 difference in peak mid-foot dorsiflexion between running conditions (ES =0.06,  $P = 0.63$ ,

349 Table 2). The mid-foot was in a position of greater plantar flexion at toe-off when participants  
350 ran with a forefoot strike (ES = 0.9,  $P \leq 0.05$ , Table 2).

351

### 352 *Joint kinetics*

353 Peak ankle plantar flexion moment was greater when running with a forefoot strike pattern  
354 (ES = 1.3,  $P \leq 0.05$ , Table 3 and Figure 3). Peak negative ankle power (ES = 2.4,  $P \leq 0.05$ ,  
355 Table 2) and positive ankle power (ES = 0.74,  $P \leq 0.05$ , Table 2) were also greater when  
356 running with a forefoot strike technique. A greater amount of negative work (ES = 2.1,  $P \leq$   
357 0.05, Table 2) and positive work (ES = 1.0,  $P \leq 0.05$ , respectively) was performed at the ankle  
358 when running with a forefoot strike.

359

360 For both running conditions, peak rear-foot plantarflexion moment occurred in mid-stance

361 (Figure 3). Peak rear-foot plantar flexion moment was greater when running with a forefoot  
362 technique (ES= 1.3,  $P \leq 0.05$ , Table 3 and Figure 3). Peak negative power was greater when  
363 running with a forefoot technique (ES = 1.6,  $P \leq 0.05$ , Table 3), however there was no  
364 difference in peak positive power between running conditions (ES = 0.3,  $P = 0.13$ , Table 3). A  
365 greater amount of negative work (ES = 1.8,  $P \leq 0.05$ , Table 3) and positive work (ES = 0.6,  $P$   
366  $\leq 0.05$ , Table 3) was performed at the rear-foot in the forefoot running condition.

367

368 In a similar manner to the ankle and rear-foot, peak mid-foot plantar flexion moment occurred

369 in mid-stance for both running conditions (Figure 3). Peak mid-foot plantar flexion moment  
370 was greater when running with a forefoot technique (ES = 1.1,  $P \leq 0.05$ , Table 3 and Figure  
371 3). Peak negative mid-foot power (ES = 2.7,  $P \leq 0.05$ , Table 3) and positive mid-foot power  
372 (ES = 0.77,  $P \leq 0.05$ , Table 3) was also greater when running with a forefoot strike. A greater  
373 amount of negative work (ES = 2.2,  $P \leq 0.05$ ) and positive work (ES = 1.1,  $P \leq 0.05$ , Table 3)

374 was also performed at the mid-foot when running with a forefoot technique, suggesting that a  
375 greater magnitude of mechanical energy is absorbed, returned and/or generated about the mid-  
376 foot. Despite similar peak deformation and an increased plantar flexion moment, the mid-foot  
377 was less stiff (more compliant) when running with a forefoot technique (ES = 0.65,  $P \leq 0.05$ ,  
378 Table 3).

379

### 380 *Muscle activation*

381 The AH muscle displayed a large burst of activity that commenced during late swing-phase,  
382 and continued through to toe-off, with peak activation occurring in mid-stance (Figure 4).

383 When running with a forefoot strike, mean AH activity was greater in swing (ES = 0.67,  $P \leq$   
384  $0.05$ , Table 4) and stance (ES = 0.63,  $P \leq 0.05$ , Table 4) phase. Peak stance phase muscle  
385 activation was also greater when running with a forefoot strike (ES = 0.71,  $P \leq 0.05$ , Table 4).

386

387 The FDB activation pattern was characterised by a large burst of activity commencing at foot  
388 contact, with a mid-stance peak and de-activation shortly after toe-off (Figure 4). As this  
389 muscle is primarily active in stance phase, mean swing phase activation was similar between  
390 conditions (ES = 0.01,  $P = 0.95$ , Table 4). Mean FDB stance phase activation substantially  
391 higher when running with a forefoot strike (ES = 0.56,  $P \leq 0.05$ , Table 4), however there was  
392 no significant difference in peak stance phase activation for this muscle (ES = 0.49,  $P = 0.19$ ,  
393 Table 4).

394

395 For both conditions, MG displayed a large burst of activation commencing in late swing,  
396 continuing through stance, with de-activation occurring during late stance-phase (Figure 4).

397 Mean swing phase activation was substantially greater when running with a forefoot strike



398 (ES = 2.36,  $P \leq 0.05$ , Table 4) while mean (ES = 0.57,  $P = 0.09$ , Table 4) and peak (ES =  
399 0.06,  $P = 0.84$ , Table 4) stance phase activity was similar for both running conditions.

400

401 The SOL activation pattern was similar for both running conditions, with a single burst of  
402 activity that commenced in terminal swing phase, peaked during mid-stance and ceased prior  
403 to toe-off (Figure 4). Swing phase activation was comparable for both running techniques (ES  
404 = 0.2,  $P = 0.39$ ), as was mean (ES = 0.50,  $P = 0.19$ ) and peak (ES = 0.69,  $P = 0.21$ ) stance  
405 phase activity (Table 4).

406

## 407 **Discussion**

408 This study provides novel evidence for active alterations to the mechanical function of the foot  
409 when [acutely](#) switching between non-forefoot and forefoot running techniques. When running at  
410 matched velocities, forefoot running resulted in higher mid-foot joint moments, more positive and  
411 more negative work being performed about the mid-foot. Increases in joint moments for forefoot  
412 striking occurred concurrently with increased intrinsic foot muscle activation and no increase in  
413 peak mid-foot dorsiflexion. Thus, it seems that the intrinsic foot muscles may be activating to  
414 [stiffen the foot during mid-stance when running with a forefoot technique](#).

415

### 416 *LA kinematics*

417 Peak LA deformation was not influenced by foot-strike technique, despite the increased joint  
418 moments about the mid-foot when running on the forefoot. This finding is analogous to a number  
419 of previous studies that also reported no difference in peak LA deformation when running with  
420 rear-foot or forefoot running techniques (30, 33, 39). Interestingly, the previous studies by Wager  
421 & Challis (2016) and McDonald et al (2016) have made the assumption in their modelling that the  
422 foot is a passive structure with no muscular control of LA stiffness. The findings of our current

423 study highlight that the foot is not passive and the plantar aponeurosis is not the sole structure  
424 responsible for maintaining LA stiffness during running. If the LA compression was indeed  
425 controlled purely by passive elastic structures (e.g. plantar aponeurosis), the increased torques  
426 observed when running with a forefoot technique would produce an increase in LA deformation  
427 (12, 22). This was not the case in the current study, or in the previously mentioned studies (30, 33,  
428 39), indicating that active muscular control must be contributing to control of LA mechanics when  
429 switching between foot-strike techniques.

430  
431 Recently published studies performed within our laboratory have highlighted the importance of the  
432 plantar intrinsic foot muscles in actively controlling the LA during quiet stance and locomotion  
433 (19-21). When activated, these muscles have the capacity to provide buttressing support for the  
434 LA, resisting excessive deformation that may occur at loads exceeding body weight (18).  
435 Furthermore, it appears that the central nervous system (CNS) can vary the relative contribution of  
436 these muscles in controlling the LA, depending on the magnitude of forces encountered when  
437 running (20). Based on our knowledge of the function of the intrinsic foot muscles (Kelly,  
438 Cresswell, Racinais, Whiteley, & Lichtwark, 2014a; Kelly, Lichtwark, & Cresswell, 2014b), it  
439 becomes apparent that the observed increase in AH and FDB activation during stance may be  
440 responsible for the lack of difference in peak LA deformation between running techniques, despite  
441 obvious differences in joint moments. An increase in force output from the intrinsic foot muscles  
442 when running with a forefoot technique may serve to reinforce the LA in mid-stance, preventing  
443 excessive LA deformation and any associated increase in plantar aponeurosis strain. Recently  
444 McDonald et al. (2016) suggested that the risk of injury to the plantar aponeurosis (i.e. plantar  
445 fasciopathy) might not be influenced by foot-strike technique, due to the similarity in peak plantar  
446 aponeurosis strains between techniques. The current study reveals that there is a substantial  
447 increase in loading and energy absorbed about the mid-foot when running with a forefoot technique

448 and that this increase is likely associated with an increase in force output from the plantar intrinsic  
449 foot muscles. Therefore, without specific training, habitual rear-foot runners may not have the  
450 strength or endurance in the intrinsic foot muscles to sustain a forefoot running technique for  
451 prolonged periods. This may subsequently influence the loading on the plantar aponeurosis. Further  
452 investigations may now be warranted to determine the time required sufficiently strengthen the  
453 intrinsic foot muscles, in order to safely transition to a forefoot running technique.

454  
455 Within the current study we observed a reduction LA stiffness when participants ran with a  
456 converted forefoot technique, despite an increase in mid-foot joint moment and no change in peak  
457 mid-foot deformation. This reduction in stiffness appears to be due to an increase in mid-foot  
458 plantar-flexion (LA height) at foot contact, allowing the mid-foot to move through a greater total  
459 range of motion in the forefoot running condition. We observed an increase AH activation in late  
460 swing, preceding an increase in LA height at foot contact when participants ran with a forefoot  
461 technique. Thus it appears that late swing activation of AH may increase the height of the LA prior  
462 to foot contact. An increase in total excursion of the LA and the associated reduction in LA stiffness  
463 may enhance the capacity for energy storage and return within this structure, without the potentially  
464 harmful effect of increased plantar fascia strain. It must however be noted that due to the lack of  
465 difference in peak LA deformation between running techniques, any increase in mechanical work  
466 performed about the LA is likely to be performed by active muscle rather than passive elastic  
467 tissues. Therefore any increase in energy absorption and return may come with increased metabolic  
468 cost.

469  
470 *Joint Moments*  
471 Forefoot running resulted in higher plantar flexion moments being generated about the ankle and  
472 mid-foot. When running with a forefoot technique, the resultant GRF is generally positioned within

473 the anterior portion of the foot, as the forefoot makes initial contact with the ground. Conversely,  
474 with a rear-foot technique the COP is initially located on the lateral aspect of the heel and  
475 progresses forward as the body progresses over the foot (6, 7, 32). If the COP is located further  
476 anterior when running with a forefoot technique, it will have a greater moment arm about the ankle  
477 and mid-foot, which, alongside an increase in peak vertical GRF, likely explains the increase in  
478 plantar flexion moments that are generated about these joints. While the finding of an increase in  
479 plantar flexion moment about the mid-foot is quite novel, a similar increase in ankle joint moment  
480 has been reported in a number of previous studies examining differences in ankle joint kinetics  
481 when switching between rear-foot and forefoot running techniques (23, 35).

482

#### 483 *Mechanical work and power*

484 The energetic benefits of the spring-like function of the LA during running have been well  
485 documented, with this mechanism providing between 8 – 17% of the mechanical energy required  
486 for each stride via the elastic stretch and recoil of the plantar aponeurosis (22, 36). Previous studies  
487 exploring the influence of foot-strike technique on elastic energy storage in the LA (30, 39) have  
488 reported that a forefoot running technique does not promote increased utilization of the elastic  
489 energy, due to the lack of difference in peak plantar aponeurosis strain between running techniques.  
490 Within our current experiment we observed an increase in mechanical power absorption and  
491 generation and an associated increase in negative and positive work about the LA, in the absence  
492 of any additional increase in LA deformation. The plantar intrinsic foot muscles are known to  
493 actively lengthen and shorten during the stance phase of running and therefore have the potential  
494 to absorb and generate mechanical power about the LA (20). The increase in intrinsic foot muscle  
495 activation when running with a forefoot technique indicates that the contribution of these muscles  
496 to mechanical work about the LA increases when running with a forefoot technique. The intrinsic  
497 foot muscles have relatively long tendons and short muscle fibres (24) and thus may be well suited

498 to elastic energy storage and return. The additional negative work done at the mid-foot joint when  
499 running with a forefoot technique may allow greater utilization of the foot spring mechanism,  
500 allowing additional energy to be stored in the tendons of the intrinsic foot muscles, without the  
501 potentially detrimental effects of excessive strain on the plantar aponeurosis. In this sense we might  
502 consider the intrinsic foot muscles as reserve actuators, able to act in parallel with the plantar  
503 aponeurosis when a certain level of force generation is required. However, it must also be noted  
504 that the increase in work performed about the LA when running with a forefoot technique may not  
505 translate directly into metabolic energy savings, due to the added energetic cost of muscle  
506 activation required to perform this work. The increased cost of activation in the intrinsic foot  
507 muscles and also the posterior leg muscles when running with a forefoot technique may in part  
508 explain the lack of difference in energetic cost between habitual rear-foot and forefoot runners (4,  
509 13) as the increased utilization of spring-like running mechanics may be offset by the additional  
510 energetic cost of muscle activation.

511  
512 Runners with an habitual forefoot technique generally display shorter ground contact times, shorter  
513 stride length and an increased cadence, compared to runners who are habitual rear-foot strikers  
514 (Gruber, Umberger, Braun, & HAMILL, 2013; Lieberman et al., 2010; Stearne, Alderson, GREEN,  
515 DONNELLY, & Rubenson, 2014). In the current experiment, our group of habitual rear-foot  
516 runners maintained relatively constant stride temporal characteristics when acutely changing to a  
517 forefoot technique. This finding is in line with a number of previous studies that have shown a lack  
518 of immediate change in stride temporal characteristics, when participants shifted from an habitual  
519 rear-foot technique to an imposed forefoot technique (Almeida, Davis, & Lopes, 2015). Thus, it is  
520 apparent that changes in stride temporal characteristics do not occur automatically when changing  
521 foot-strike technique{papers2\_bibliography}. Specific training to encourage a shorter stride length

522 and increased cadence, may also need to be included as part of a habituation process in a transition  
523 to forefoot running.

524

525 *Methodological considerations*

526 The LA is a multi-articular structure (17). We have modelled this structure as single functional  
527 mid-foot joint. This technique allows important general insights into the function of the LA,  
528 however, it may also be an over-simplification of the complex sequence of rotations and  
529 translations that occur about a series of small joints within this overall structure. Furthermore, we  
530 have only modelled the function of this structure in the sagittal plane. It is likely that motion of the  
531 LA also occurs in the frontal and transverse planes (Kelly et al., 2014a; Leardini et al., 2007).

532

533 The use of an inverse dynamic solution allowed us to provide unique insight to the mechanical  
534 function of the LA when running with rear-foot and forefoot running techniques. A limitation of  
535 this approach is that it is difficult to quantify the relative percentage of GRF that is acting on the  
536 forefoot segment at any given time in stance, as some force may be simultaneously applied to the  
537 rear-foot. However, as the COP rapidly progresses anteriorly in stance phase and the forefoot bears  
538 a substantial portion of the GRF force for the majority of stance duration (6, 7), any errors  
539 associated with this assumption are likely to be small and localized to the initial period of stance  
540 in the habitual running condition. In order to mitigate this limitation, we have calculated rear-foot  
541 and mid-foot moments and powers after the COP has progressed anterior to the axis of the mid-  
542 foot joint. The COP passed the axis of the mid-foot at approximately 25% stance phase in the  
543 habitual running condition and after this time we are confident that the vast majority of contact  
544 forces are applied to the forefoot. When running with a forefoot running technique the COP was  
545 anterior to the axis of the mid-foot joint for the entire duration of stance phase, therefore the GRF  
546 is applied to the forefoot segment from the instant of foot contact. If small errors do exist based on

547 our GRF assignment assumption, these are limited to the rear-foot running condition and will lead  
548 to an over-estimation in LA joint moments, therefore the magnitude of differences in joint kinetics  
549 between conditions may actually be slightly larger than those reported. Providing further support  
550 to the validity of our inverse dynamic approach, the joint moments at the rear-foot are comparable  
551 to the ankle joint moments using a traditional rigid foot segment, both within our current study and  
552 also to the values presented in previous studies reporting ankle joint moments during running at  
553 similar speeds (34).

554  
555 Participants within this experiment were shod rear-foot runners who were instructed to run with  
556 their habitual foot-strike technique and a forefoot technique. It is possible that different  
557 observations may have been made if habitual forefoot runners were included in this experiment.  
558 However based on the findings of previous studies (30, 35) it is unlikely that LA kinematics would  
559 differ between habitual rear-foot and forefoot runners when instructed to run with forefoot and  
560 rear-foot techniques. Furthermore, given that activation of the intrinsic foot muscles is dependent  
561 on loading requirement (18), it is unlikely that the activation patterns of these muscles would  
562 substantially differ either.

563  
564 **Conclusion**  
565 Foot-strike technique substantially influences the mechanical function of the LA during running.  
566 A forefoot running technique leads to an increase in loading about the mid-foot and an increase in  
567 positive and negative mechanical work performed by the intrinsic foot muscles. These findings  
568 suggest that adequate strength and endurance training for the intrinsic foot muscles should be  
569 included as part of training programs for runners seeking to transition from a rear-foot to a forefoot  
570 running technique.

571

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577

578 **Conflict of interest**

579 The authors declare no conflict of interest.

580

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584

585

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690 **Figure 1.** Depiction of sagittal plane angular rotations about the Ankle (top), Rear-foot  
691 (middle) and Mid-foot (bottom) joints. Angles in the text are relative to the respective joint  
692 angle in quiet stance, with  $0^{\circ}$  representing the joint angle in quiet stance. Negative angles  
693 denote dorsiflexion (upwards rotation of the distal segment).

694 **Figure 2.** Group mean  $\pm$  standard deviation (shaded area) for vertical ground reaction force  
695 (GRF, top) and anterior-posterior GRF (bottom). Data is recorded from each participant  
696 running with a habitual (solid) and forefoot running technique (dashed) at 2.8 ms<sup>-1</sup>. Data are  
697 shown for stance only; from right foot contact (FC) to right toe off (TO). All data are  
698 normalised to body weight (BW). \* denotes a significant difference in peak vertical GRF  
699 between the running conditions.

700 **Figure 3.** Group mean  $\pm$  standard deviation (shaded area) for joint angle (top), normalised  
701 joint moment (middle) and normalised joint power (bottom) calculated during stance phase  
702 for the mid-foot (left), rear-foot (middle) and ankle (right). Data is recorded from each  
703 participant running with a habitual (solid) and forefoot running technique (dashed) at 2.8 ms-  
704 1. Data are shown for stance only; from right foot contact (FC) to right toe off (TO). Vertical  
705 dashed lines denote the time point that the centre of pressure progressed anterior to the axis of  
706 rotation of the mid-foot joint in the habitual foot-strike condition. Joint work is calculated by  
707 trapezoidal integration of the joint power data.  $\delta$  denotes a significant difference between  
708 running conditions in joint angle (at FC, peak or TO) for each joint.  $\gamma$  denotes significant  
709 difference in peak joint moment between the running conditions.  $\beta$  denotes a significant  
710 difference in peak joint power (negative or positive) and  $\alpha$  denotes a significant difference in  
711 negative or positive joint work between running conditions

712 **Figure 4.** Group mean ensembles  $\pm$  standard deviation (shaded area) for electromyography  
713 (EMG) normalised root mean square signal amplitude for medial gastrocnemius (MG), soleus  
714 (SOL), flexor digitorum brevis (FDB) and abductor hallucis (AH). Group mean ensembles are  
715 defined from toe off (TO) to ipsilateral toe off for the right foot. Data recorded during running  
716 at 2.8 ms<sup>-1</sup>. For each muscle EMG data is normalised to the maximal amplitude recorded for  
717 all trials. The habitual foot-strike condition is the solid lines and the forefoot condition the  
718 dashed lines. Vertical lines denote onset of stance phase at foot contact (FC). \* denotes  
719 significant difference between running conditions.

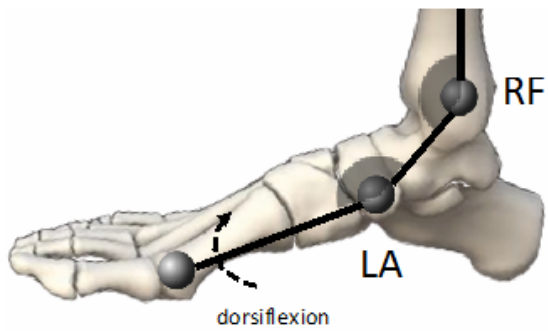
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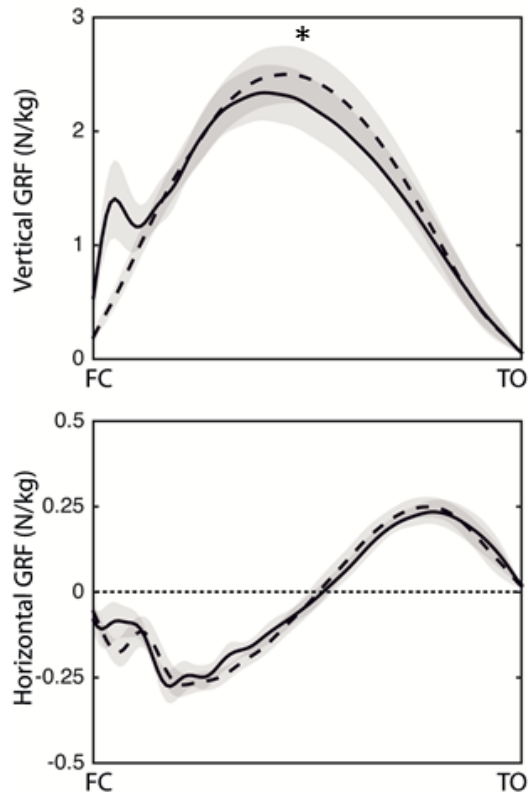


**Figure 1.** Depiction of the Lower limb marker set employed for collection of kinematic data. Knee markers are removed following a static calibration trial, with the cluster of four markers on a rigid plastic shell used to track the motion of the shank. The posterior calcaneus marker cannot be viewed in this image.

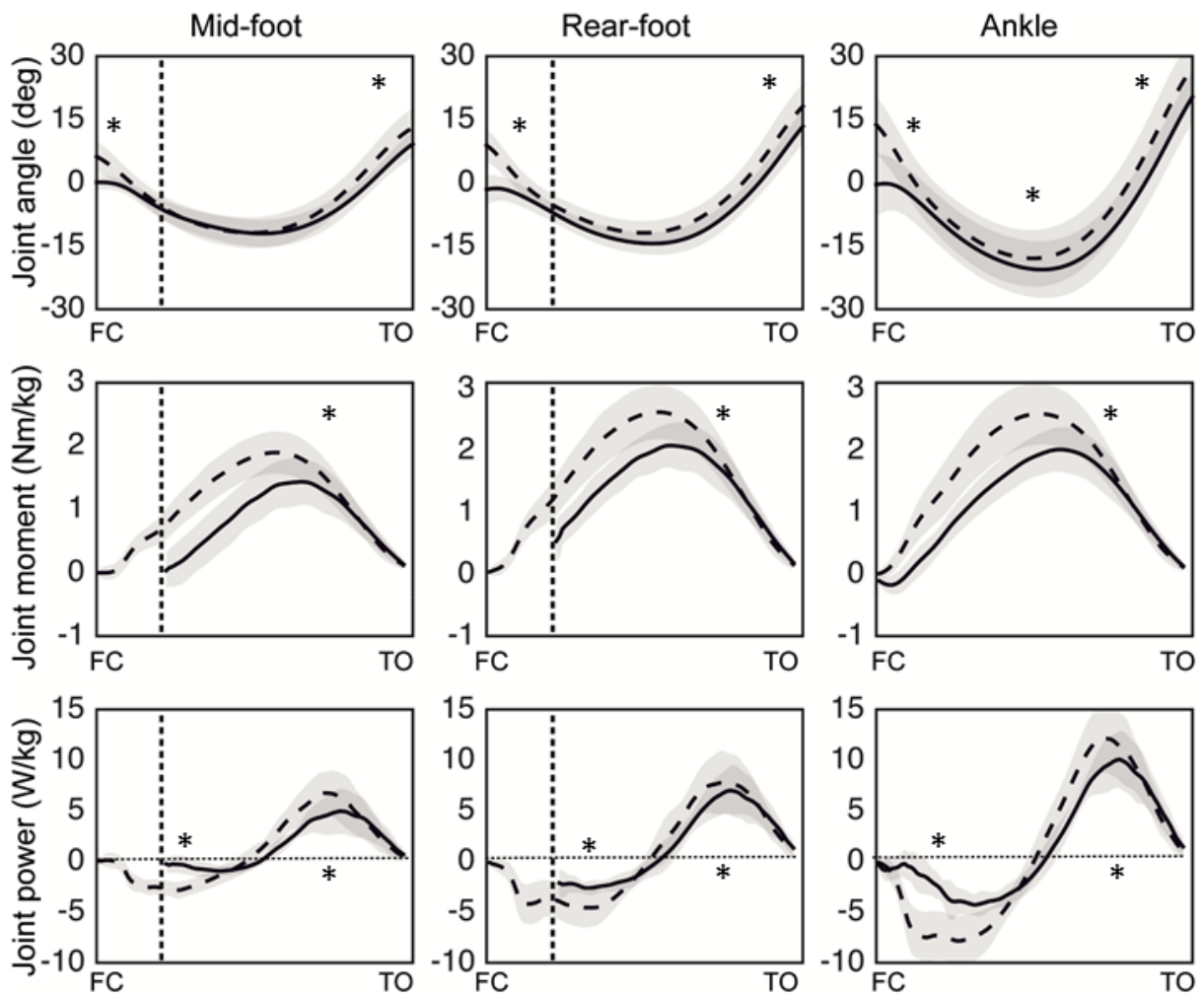


**Figure 2.** Depiction of angular rotations about the longitudinal arch (LA) and rear-foot (RF) joints. Positive angles denote plantar flexion. Increase in LA angle indicates an increase in height of the LA.

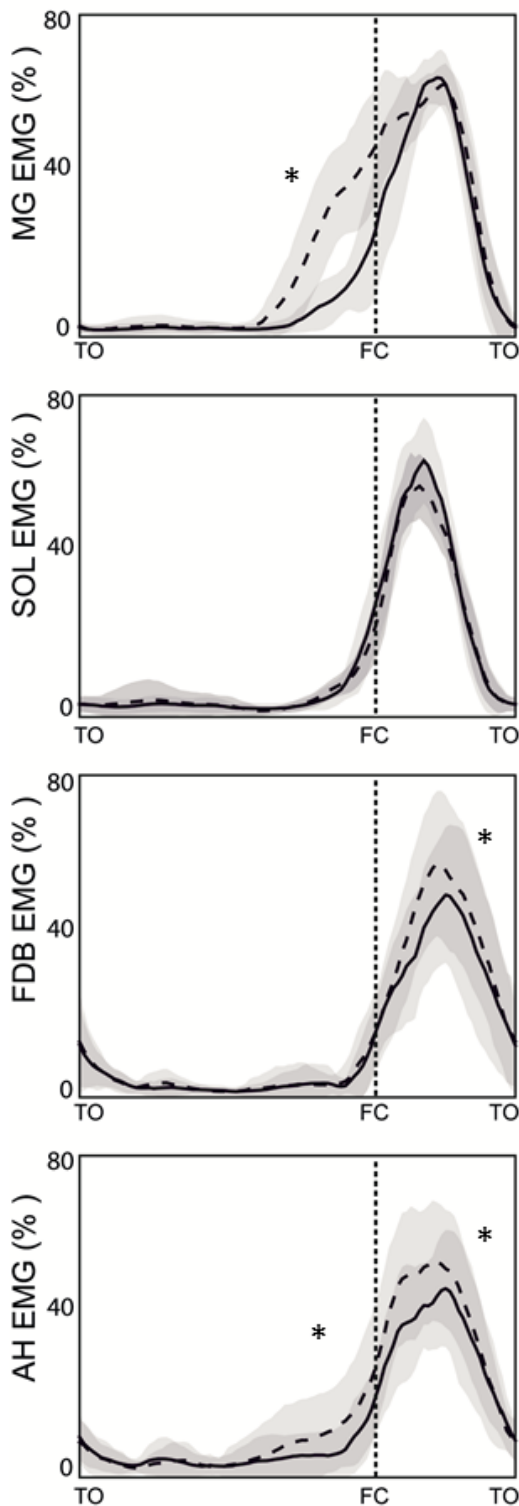




**Figure 3.** Group mean  $\pm$  standard deviation (shaded area) for vertical force (top) and antero-posterior force (bottom). Data is recorded from each participant running with a preferred (solid) and forefoot running technique (dashed) at  $2.8 \text{ ms}^{-1}$  and presented from from foot contact to toe off from the right foot. All data is normalised to body mass.



(bottom) calculated during stance phase for the longitudinal arch (left), rear-foot (middle) and ankle (right). Data is recorded from each participant running with a preferred (solid) and forefoot running technique (dashed) at  $2.8 \text{ ms}^{-1}$ . All data is normalised to body mass. Vertical dashed lines denote initial forefoot loading in preferred foot-strike condition.



**Figure 5.** Group mean ensembles  $\pm$  standard deviation (shaded area) for electromyography (EMG) normalised root mean square signal amplitude for gastrocnemius medialis (MG), soleus (SOL), flexor digitorum brevis (FDB) and abductor hallucis (AH). Group mean ensembles are defined from toe off (TO) to ipsilateral toe off for the right foot. Data recorded during running at  $2.8 \text{ ms}^{-1}$ . For each muscle EMG data is normalised to the maximal amplitude recorded for all trials. The preferred foot-strike condition is the solid lines and the forefoot condition the dashed lines. Vertical lines denote onset of stance phase.

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