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

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Abstract

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

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

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

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

Keywords

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

51

Introduction

The orientation of a runner's foot at the moment of foot-ground impact is often employed as a

method to classify running technique (2, 7, 29, 33). Foot-strike patterns are generally classified

in three groups; (1) runners who land in a heel-first manner (rear-foot strike), (2) runners who

land on the forefoot (forefoot strike), and (3) runners who land with a simultaneous heel and

forefoot contact (mid-foot strike) (3, 7). Although the majority (>75%) of distance runners

adopt a rear-foot running technique (14, 25), it has been suggested that a forefoot running

technique is mechanically advantageous, as it affords the runner greater opportunity to recycle

the energy associated with foot-ground impact via elastic stretch and recoil of the tendons and

ligaments of the ankle and foot (28, 29, 33). Consequently, the popularity of running "re-

training" programs that promote a forefoot landing pattern to reduce injury risk and improve

running performance have surged (3, 10, 15).

The longitudinal arch (LA) of the human foot behaves in a spring-like manner when running,

temporarily storing and then subsequently returning a considerable portion of the mechanical

energy required for each stride (22, 36, 39). Compression (lowering and lengthening) of the LA

during the first half of stance phase allows mechanical energy to be temporarily stored within

the stretched ligaments, muscles and tendons that span this structure (20, 22, 39). This energy

is subsequently returned in late stance, as the resultant ground reaction force (GRF) declines

and the stretched elastic structures shorten to allow the LA to recoil (rise and shorten) (20, 22,

33).

Recently studies have explored the hypothesis that a forefoot running technique enhances foot-

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 ± 6 years; height: 172 ± 8

114 cm; mass: 72 ± 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 m.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,

joint moments for the rear-foot and mid-foot were only calculated during stance phase after the

COP had progressed distal to the axis of the mid-foot joint, minimising any potential errors

associated with segment force assignments (37). Subsequently, joint moments and joint powers

for the ankle, rear-foot and mid-foot were filtered using a 20 Hz recursive, second order low-

pass Butterworth filter (Edwards, Troy, & Derrick, 2011). Net negative and positive work

performed by each joint during stance phase was calculated by applying a trapezoidal

integration of joint power curves across the stance phase. Mid-foot stiffness was quantified by

diving the change in mid-foot joint moment by the change in mid-foot angle during stance phase

(Farley & Morgenroth, 1999).

Muscle activation

All EMG signals were high-pass filtered using a recursive second order Butterworth filter at 35

Hz to remove any unwanted low-frequency movement artefact. Subsequently, EMG signals

were visually inspected in order to identify any remaining artefact, which was defined as an

abnormal spike in the signal, typically associated with foot contact. Any remaining artefacts

resulted in the EMG data for that particular stride being excluded from further analysis.

Following DC-offset removal, root mean square (RMS) signal amplitude was calculated across

all EMG data using a moving window of 50 ms to generate an EMG envelope. The EMG

envelope for each muscle was then normalized to the maximum RMS amplitude found across

all conditions for the respective muscle. The normalized peak EMG RMS envelope amplitude

was calculated during stance phase, additionally the normalized mean EMG RMS was

calculated during both the stance and swing phases, allowing comparisons in magnitude of both

stance and swing phase muscle activation between HFS and FFS conditions.

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 ≤ 0.70 representing a small

307 effect, >0.70 and <1.2 representing a moderate effect, and ≥ 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 ≤ 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 ≤ 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 ≤ 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 ≤ 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 ≤ 0.05, Table 2). The rear-foot was more plantar flexed at toe-off with the

342 forefoot running technique (ES = 1.1, P ≤ 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 ≤ 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 ≤ 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 ≤ 0.05, Table 3 and Figure 3). Peak negative ankle power (ES = 2.4, P ≤ 0.05,

355   Table 2) and positive ankle power (ES = 0.74, P ≤ 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 ≤

357   0.05, Table 2) and positive work (ES = 1.0, P ≤ 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 ≤ 0.05, Table 3 and Figure 3). Peak negative power was greater when

363   running with a forefoot technique (ES = 1.6, P ≤ 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 ≤ 0.05, Table 3) and positive work (ES = 0.6, P

366   ≤ 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 ≤ 0.05, Table 3 and Figure

371   3). Peak negative mid-foot power (ES = 2.7, P ≤ 0.05, Table 3) and positive mid-foot power

372   (ES = 0.77, P ≤ 0.05, Table 3) was also greater when running with a forefoot strike. A greater

373   amount of negative work (ES = 2.2, P ≤ 0.05) and positive work (ES = 1.1, P ≤ 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 ≤ 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 ≤

384 0.05, Table 4) and stance (ES = 0.63, P ≤ 0.05, Table 4) phase. Peak stance phase muscle

385 activation was also greater when running with a forefoot strike (ES = 0.71, P ≤ 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 ≤ 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 ≤ 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. 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

17

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

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

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

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

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586   **References**

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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° representing the joint angle in quiet stance. Negative angles

693 denote dorsiflexion (upwards rotation of the distal segment).

694 **Figure 2.** Group mean ± 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-1. 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 ± 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. δ denotes a significant difference between

708 running conditions in joint angle (at FC, peak or TO) for each joint. γ denotes significant

709 difference in peak joint moment between the running conditions. β denotes a significant

710 difference in peak joint power (negative or positive) and α denotes a significant difference in

711 negative or positive joint work between running conditions

712 **Figure 4.** Group mean ensembles ± 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^{-1}. 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.
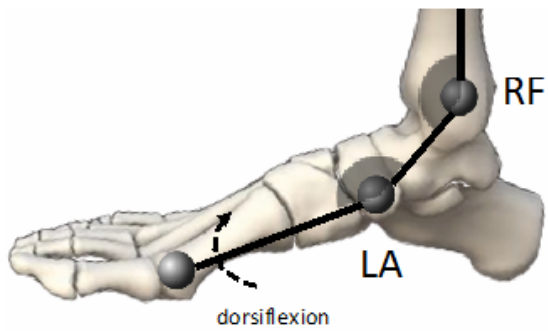
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Figure 1. Depiction of the Lower limb marker set employed for collection of kinematic data. Knee markers are removed following a static calibration trial, with the cluster of four markers on a rigid plastic shell used to track the motion of the shank. The posterior calcaneus marker cannot be viewed in this image.

723

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

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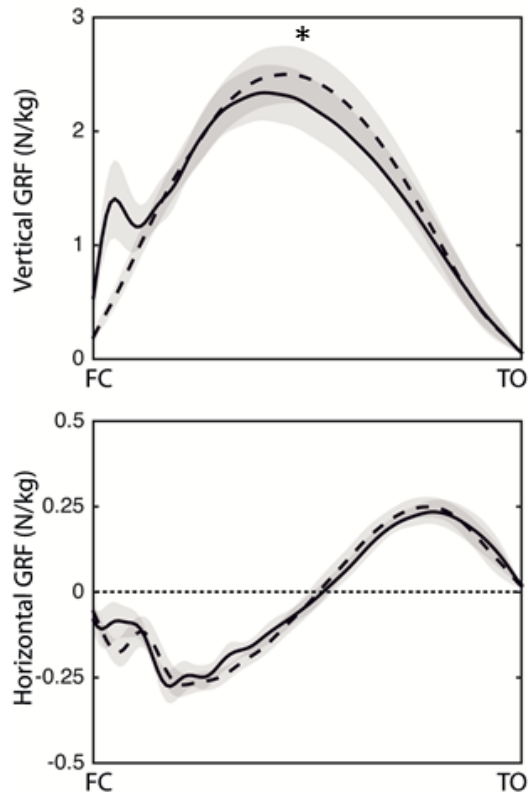
Figure 3. Group mean ± standard deviation (shaded area) for vertical force (top) and antero-posterior force (bottom). Data is recorded from each participant running with a preferred (solid) and forefoot running technique (dashed) at 2.8 ms^{-1} and presented from from foot contact to toe off from the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

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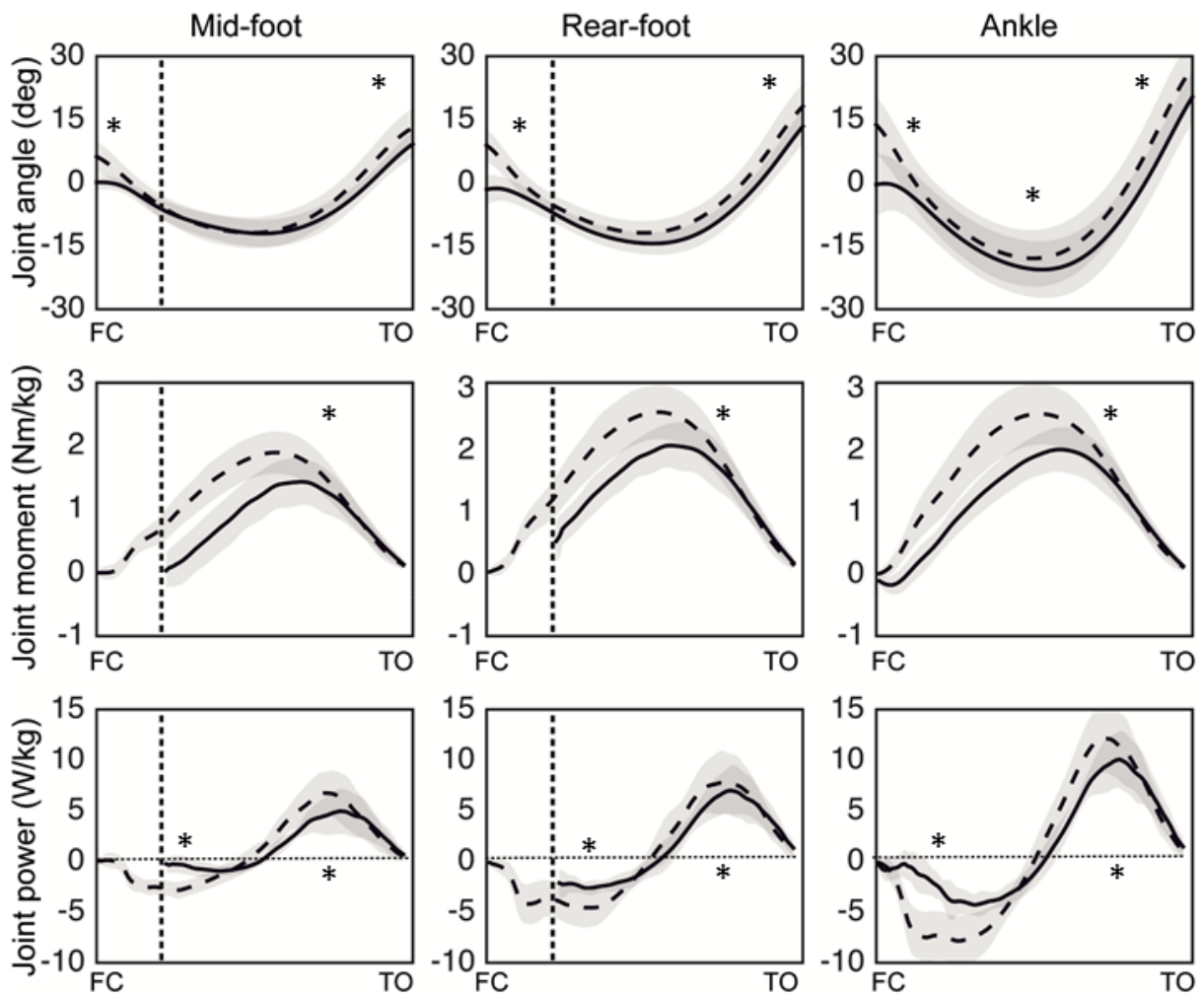
(bottom) calculated during stance phase for the longitudinal arch (left), rear-foot (middle) and ankle (right). Data is recorded from each participant running with a preferred (solid) and forefoot running technique (dashed) at 2.8 ms^{-1}. All data is normalised to body mass. Vertical dashed lines denote initial forefoot loading in preferred foot-strike condition.
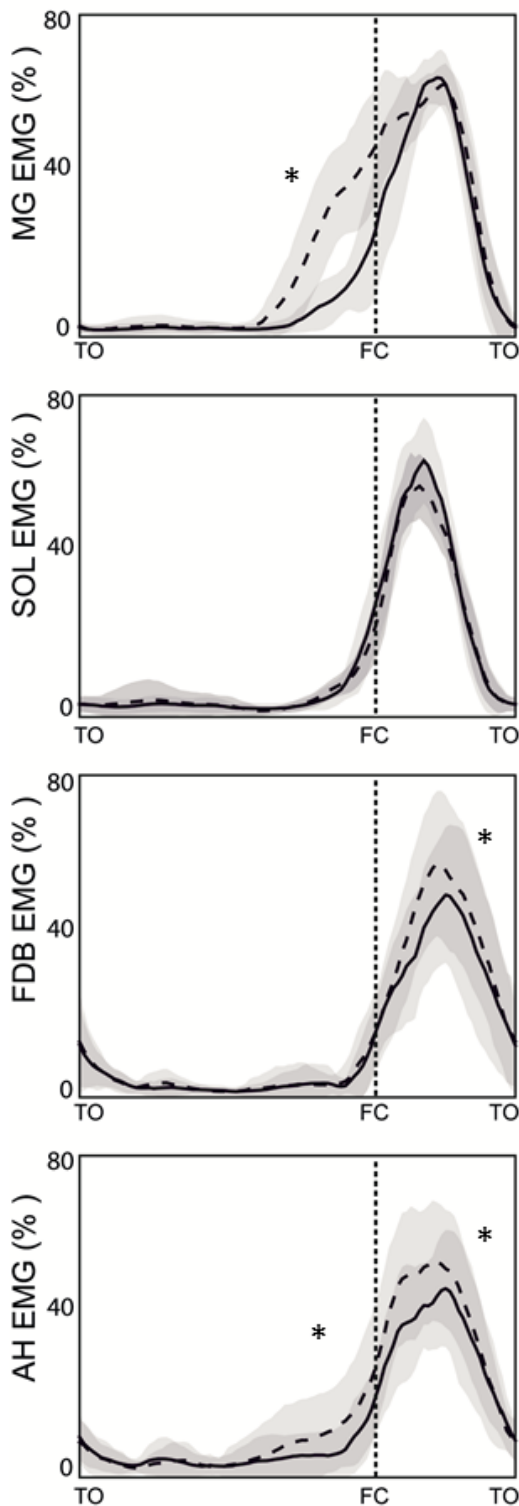
Figure 5. Group mean ensembles ± standard deviation (shaded area) for electromyography (EMG) normalised root mean square signal amplitude for gastrocnemius medilais (MG), soleus (SOL), flexor digitorum brevis (FDB) and abductor hallucis (AH). Group mean ensembles are defined from toe off (TO) to ipsilateral toe off for the right foot. Data recorded during running at 2.8 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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