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Title: The biomechanical characteristics of wearing FitFlop™ sandals highlight significant alterations in gait pattern: A comparative study.

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Corresponding Author: Dr. Darren C James,

Corresponding Author's Institution: London South Bank University

First Author: Darren C James

Order of Authors: Darren C James; Laura J Farmer; Jason B Sayers; David P Cook; Katya N Mileva

Abstract: Background: The net contribution of all muscles that act about a joint can be represented as an internal joint moment profile. This approach may be advantageous when studying footwear-induced perturbations during walking since the contribution of the smaller deeper muscles that cross the ankle joint cannot be evaluated with surface electromyography. Therefore, the present study aimed to advance the understanding of FitFlop™ footwear interaction by investigating lower extremity joint moment, and kinematic and centre of pressure profiles during gait.

Methods: 28 healthy participants performed 5 walking trials in 3 conditions: a FitFlop™ sandal, a conventional sandal and an athletic trainer. Three-dimensional ankle joint, and sagittal plane knee and hip joint moments, as well as corresponding kinematics and centre of pressure trajectories were evaluated.

Findings: FitFlop™ differed significantly to both the conventional sandal and athletic trainer in: average anterior position of centre of pressure trajectory ($P < 0.0001$) and peak hip extensor moment ($P = 0.001$) during early stance; average medial position of centre of pressure trajectory during late stance; peak ankle dorsiflexion and corresponding range of motion; peak plantarflexor moment and total negative work performed at the ankle (all $P < 0.0001$).

Interpretation: The present findings demonstrate that FitFlop™ footwear significantly alters the gait pattern of wearers. An anterior displacement of the centre of pressure trajectory during early stance is the primary response to the destabilising effect of the mid-sole technology, and this leads to reductions in sagittal plane ankle joint range of motion and corresponding kinetics. Future investigations should consider the clinical implications of these findings.

Highlights

- Gait was investigated in people wearing FitFlop™ footwear.
- An early change in centre of pressure location is the primary response.
- The sagittal plane ankle moment and range of motion are reduced throughout stance.
- These findings may have clinical relevance.

The biomechanical characteristics of wearing FitFlop™ sandals highlight significant alterations in gait pattern: A comparative study.

Darren C. James ^a, Laura J. Farmer ^a, Jason B. Sayers ^a, David P. Cook ^a, Katya N. Mileva ^a

^a Sport & Exercise Science Research Centre, School of Applied Sciences, London South Bank University, London, UK.

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Corresponding author:

Dr Darren James

School of Applied Sciences, ESBE

London South Bank University

103 Borough Road, London, SE1 0AA, UK.

Email: jamesd6@lsbu.ac.uk

TEL: +44 (0)207 815 7992

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1 Abstract

2 *Background:* The net contribution of all muscles that act about a joint can be represented as an
3 internal joint moment profile. This approach may be advantageous when studying footwear-
4 induced perturbations during walking since the contribution of the smaller deeper muscles that
5 cross the ankle joint cannot be evaluated with surface electromyography. Therefore, the present
6 study aimed to advance the understanding of FitFlop™ footwear interaction by investigating
7 lower extremity joint moment, and kinematic and centre of pressure profiles during gait.

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9 conventional sandal and an athletic trainer. Three-dimensional ankle joint, and sagittal plane
10 knee and hip joint moments, as well as corresponding kinematics and centre of pressure
11 trajectories were evaluated.

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16 plantarflexor moment and total negative work performed at the ankle (all $P<0.0001$).

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18 gait pattern of wearers. An anterior displacement of the centre of pressure trajectory during early
19 stance is the primary response to the destabilising effect of the mid-sole technology, and this
20 leads to reductions in sagittal plane ankle joint range of motion and corresponding kinetics.
21 Future investigations should consider the clinical implications of these findings.

1 **1. Introduction**

2 Technology-inspired footwear aims to offer advantages in sporting performance,
3 restore ‘natural’ foot function and promote well-being, as well as assist mobility in a number
4 of pathological conditions. In the last decade instability shoes have come to the fore under the
5 premise that lower extremity muscles can be trained when the musculoskeletal system is
6 functionally destabilised by a midsole-induced perturbation (Nigg et al., 2006). In principle,
7 this concept is well-founded since traditional balance training is known to induce
8 sensorimotor adaptations that result in spinal and supraspinal neural reorganisation (Taube et
9 al., 2008). Accordingly, balance training strategies are used not just for rehabilitation but also
10 for improving muscle performance by stressing the musculotendinous system (Taube et al.,
11 2008).

12 However, the evidence that instability shoes enhance muscle activation profiles
13 during walking when compared to a control shoe is equivocal. Indeed, some studies have
14 shown Masai Barefoot Technology[®] (MBT[®]), the most notable unstable shoe concept, to
15 significantly increase muscle (m.) gastrocnemius activation amplitude during loading
16 response of the gait cycle when compared to conventional footwear (Price et al., 2013;
17 Romkes et al., 2006); whereas others have not (Branthwaite et al., 2013; Nigg et al., 2006).
18 Similar observations have been demonstrated in muscles from the quadriceps group
19 (Branthwaite et al., 2013; Nigg et al., 2006; Price et al., 2013; Romkes et al., 2006) and in m.
20 peroneus longus activity (Branthwaite et al., 2013; Price et al., 2013) throughout the gait
21 cycle. Despite this lack of consensus, there is increasing belief, substantiated from findings
22 on static balance control (Coza et al., 2009; Landry et al., 2010), that unstable footwear
23 activates the smaller muscles crossing the ankle joint more so than conventional or athletic
24 footwear (Burgess and Swinton, 2012; Maffiuletti, 2012; Nigg et al., 2012).

25 The FitFlop™ sandal is an innovative form of unstable footwear. The concept
26 underpinning this footwear, Microwobbleboard™ technology, is a column based triple density
27 midsole design (Fig. 1A) intended to induce a movement strategy in such a way that
28 facilitates the second ankle rocker process (Perry and Burnfield, 2010). In principle, this
29 should evoke enhanced activity from the stabilising leg muscles through the moderate medio-
30 lateral (M-L) destabilising effect afforded by the midsole construction. FitFlop™ interaction
31 has been reported to effectively apply frontal plane instability (Price et al., 2013); however,
32 there is no published evidence so far of enhanced muscle activation profiles unique to this
33 midsole technology (Burgess and Swinton, 2012; Price et al., 2013).

34 All extrinsic muscles that cross the ankle joint are potentially involved in controlling
35 perturbations during gait (Nigg et al., 2012), but the contribution from all participating
36 muscles cannot be evaluated due to the inherent difficulty associated with the acquisition of
37 electromyographic (EMG) activation profiles from the smaller, deeper-lying muscles (Kamen
38 and Caldwell, 1996). Instead, the internal joint moment profile can be used to represent the
39 net contribution of all muscles that act about a joint (Lloyd and Besier, 2003). Calculations
40 using an inverse dynamics approach, which derives a joint moment profile from movement
41 kinematics and ground reaction forces, may prove successful in understanding FitFlop™
42 interaction further. Currently, no information regarding lower extremity joint kinetics has
43 been forthcoming in the literature with respect to this footwear, and investigation is therefore
44 warranted. Consequently, the purpose of the present study was to compute and compare
45 lower extremity joint moment profiles during walking in three different types of footwear: a
46 FitFlop™ sandal, a comparative sandal and a standard athletic trainer. Joint angular
47 kinematics and centre of pressure (COP) trajectory were also assessed for differences due to
48 footwear. Based on the properties of the Microwobbleboard™ technology, and the
49 ineffectiveness of soft mid-sole constructions to produce reactive forces (Perry et al., 2007), it

50 was hypothesised that wearing FitFlop™ sandals would significantly change the anterior-
51 posterior COP trajectory during early stance, due to altered lower limb net joint moment
52 profiles. Consequently, kinematic alterations in gait were anticipated. The results from this
53 study may help to inform health practitioners of the functional adaptations imposed on the
54 wearer by the FitFlop™ sandal when prescribing technology-inspired footwear for assistive
55 mobility.

56

57 **2. Methods**

58 *2.1 Participants*

59 Twenty eight healthy individuals, 13 males (mean (SD): 28.8 (8.8) years, 78.0 (12.1)
60 kg, 1.74 (0.07) m) and 15 females (mean (SD): 31.2 (7.3) years, 64.4 (4.9) kg, 1.65 (0.04) m)
61 were informed of the testing procedures and provided written informed consent to participate
62 in the study. Prior approval was received from the local University Research Ethics
63 Committee (UREC 1021). Participants reported to be in good health and free from any recent
64 orthopaedic trauma, underlying pathology or neurological problems.

65 Sample size estimation ($P < 0.05$, $\beta = 0.20$) was based on ankle joint angular (plantarflexor)
66 impulse ($\text{Nm/Kg}\cdot\text{s}^{-1}$) data from a pilot trial investigating FitFlop™ footwear. The angular
67 impulse represents the angular moment of force acting over a specified period of time and
68 provides a useful concept for understanding loading rate (Stefanyshyn et al., 2006). In the
69 context of the present study, it allows differentiation of the impact of footwear on joint
70 energetics.

71 *2.2 Experimental design*

72 Three dimensional (3-D) lower extremity kinematics and force data were measured in
73 three conditions: a FitFlop™ Walkstar sandal (FF), a market comparative sandal
74 (Birkenstock® Gizeh; BIRK), and a standard commercially-available athletic trainer free from

75 any technological construct (Decathlon Kalenji Success, 0.39 EVA, Shore 55C, KAL)(Fig.
76 1B). Windows were cut into the trainers so that an exact representation of 3-D position data
77 of the foot segment could be collected. The testing protocol consisted of five repeated
78 walking trials in each condition at individually-preferred walking speed. This speed was
79 determined prior to the commencement of the protocol as a range for each participant to walk
80 within based on their average speed (SD 5%) from five trials recorded in the KAL condition.
81 Condition trials were randomised to exclude any potential order effect. The participants were
82 given sufficient time to familiarise walking in each condition and to establish their starting
83 position so that a right foot contact was made on an embedded force platform corresponding
84 to at least the sixth step from gait initiation. This is well beyond the time required to elicit a
85 steady state walking pattern (Couillandre and Breniere, 2003).

86

87 2.3 *Data acquisition and processing*

88 Kinematic data were acquired using an eight camera 3-D motion analysis system
89 (Oqus 3-series, Qualisys AB, Sweden), sampled at 120Hz and synchronously collected with
90 force platform data (type 9281E, Kistler, UK) at 2040Hz. Since gait in healthy subjects is
91 considered generally symmetrical at preferred walking speed (Seeley, Umberger, & Shapiro,
92 2008), only data from the right extremity were entered for statistical analysis. This approach
93 is consistent with the related literature (Burgess and Swinton, 2012; Branthwaite et al., 2013;
94 Nigg et al., 2006; Price et al., 2013; Romkes et al., 2006).

95 The 3-D pose of seven body segments of the lower extremity (pelvis; left and right
96 thighs; left and right shanks; both feet) were reconstructed by tracking the trajectories of 26
97 retro-reflective spherical markers mounted in accordance with an accepted six degree-of-
98 freedom marker set (6DOF, (Cappozzo et al., 1995). A further 18 markers were placed

99 bilaterally on anatomical landmarks during a static barefoot calibration, in order to define
100 each segment's local coordinate system (Cappozzo et al., 1995; Collins et al., 2009; Leardini
101 et al., 2007). These were subsequently removed prior to the dynamic trials so that 6DOF joint
102 movement was expressed relative to the 'calibrated anatomical systems technique' (Cappozzo
103 et al., 1995).

104

105 2.3.1 *Joint kinematics.*

106 Raw marker trajectories and ground reaction force (GRF) data were exported into
107 Visual 3D software (C-Motion Inc., USA) and smoothed with a 10Hz and 25Hz 4th order
108 low-pass Butterworth filter, respectively. Joint rotations were calculated using an X (sagittal),
109 Y (frontal), Z (transverse) Cardan rotation sequence and were referenced to coordinate
110 systems embedded in the distal segment, such that ankle dorsiflexion (DF), adduction (ADD)
111 (commonly referred to as inversion), and internal rotation (INT) were positive. Only sagittal
112 plane rotations were reported at the knee and hip joints, thus a positive rotation reflects
113 extension (KE) and flexion (HF), respectively. 3-D ankle joint range of motion (ROM, °):
114 peak plantarflexion (PF)-DF, peak abduction (ABD)-ADD, peak external rotation (EXT)-
115 INT; and sagittal plane knee (initial contact (IC)-peak knee flexion (KF)) and hip (IC-peak
116 hip extension (HE)) joint ROM, and the respective peak angles (°) were derived from stance
117 phase of the gait cycle. Stance time and step length were also extracted for statistical analysis.

118

119 2.3.2 *Joint kinetics.*

120 A Newton-Euler inverse dynamics approach was employed to calculate the 3-D
121 internal moments acting about the lower extremity joints. Again, only sagittal plane moments

122 were reported from the knee and hip joints. The moments were expressed relative to a distal
123 anatomical frame of reference and normalised to bodyweight (Nm/kg). The respective peaks,
124 times (% stance) and overall joint angular impulse (Nm/kg.s⁻¹) were derived during stance
125 phase. Also, ankle joint power, representing the sum of powers within the segment coordinate
126 system, was used to express the total negative and positive periods within the signal as an
127 indication of the total work (J.kg⁻¹) performed at the joint.

128

129 2.3.3 *Centre of pressure (COP).*

130 The COP trajectory was resolved into a virtual foot local coordinate system from
131 contact with the force platform (Visual3D, C-Motion Inc., USA). Specifically, the forward
132 progression COP was normalised (arbitrary unit, a.u) by the distance along the anterior-
133 posterior (A-P) axis from the proximal end of the foot segment (ankle joint) to the 2nd
134 metatarsal head (distal joint centre) (O'Connell et al., 1998). This meant that A-P COP range
135 of motion was quantified on the order of -1 to 2, where a negative value indicates that COP is
136 behind the ankle joint centre and a value > 1 reflects COP ahead of the metatarsals
137 (O'Connell et al., 1998). Similarly, the medio-lateral (M-L) COP was normalised (a.u) by its
138 distance along the distal radius of the foot segment (1st to 5th metatarsal head) with respect to
139 the longitudinal axis of the foot segment. An M-L COP equal to zero reflects a position
140 located on the A-P axis, whereas a positive value indicates a laterally-directed trajectory
141 (Visual3D, C-Motion Inc., USA). The data were expressed relative to subdivisions of stance
142 phase, representing early (0-33%; COP33), mid- (34-66%; COP66) and late stance (67-
143 100%; COP100) regions (Chang et al., 2008).

144

145 2.4 Statistical analysis

146 All outcome variables were determined for each of five trials completed by the
147 participant in each condition, averaged, and then compared across conditions. All data were
148 confirmed as being normally distributed (Kolmogorov-Smirnov 1-sample test, PASW v18.0,
149 IBM Corp., USA), hence a single factor (condition: FF vs BIRK vs KAL) repeated measures
150 ANOVA was used to identify main and evaluate the effect sizes (η^2). *Post-hoc* Holm-Sidak
151 corrections were applied for pair-wise comparisons and statistically significant differences
152 were accepted when $P < 0.05$.

153

154 3. Results

155 There was no significant difference in walking speed ($P > 0.05$) between conditions
156 (mean (SD); KAL: 1.45 (0.15) m/s; BIRK: 1.44 (0.15) m/s; FF: 1.44 (0.14) m/s).

157

158 3.1 Joint kinematics

159 Mean ($n=28$) joint angular kinematic profiles are presented in Figure 2. Condition
160 effects were found at the ankle and knee joints and Table 1 highlights where significant
161 differences from the pair-wise comparisons existed. At the ankle joint, there were significant
162 amplitude differences in peak plantarflexion during early stance ($F_{(2, 54)}=55.5$, $P < 0.0001$,
163 $\eta^2=0.67$), and peak dorsiflexion ($P < 0.0001$, $\eta^2=0.76$), adduction ($P < 0.0001$, $\eta^2=0.28$) and
164 internal rotation ($P=0.001$, $\eta^2=0.23$) during late stance. There were significant differences in
165 ankle ROM measured in all three planes: PF-DF ($P < 0.0001$, $\eta^2=0.49$), ABD-ADD
166 ($P < 0.0001$, $\eta^2=0.30$) and EXT-INT ($P < 0.0001$, $\eta^2=0.25$).

167 At the knee, there was a significant difference in peak knee flexion in stance
168 ($P<0.0001$, $\eta^2=0.37$) and ROM ($P=0.002$, $\eta^2=0.21$). No significant differences ($P>0.05$) were
169 found at the hip joint between conditions and comparable stance time and step length
170 measures ($P>0.05$) were observed.

171

172 3.2 Joint kinetics

173 Mean ($n=28$) joint moment ensemble profiles are presented in Figure 3. Condition
174 effects were found at all joints and Table 2 highlights where significant findings from the
175 pair-wise comparisons existed. At the ankle joint there were significant differences in the
176 peak DF moment ($P=0.001$, $\eta^2=0.23$) and time ($P<0.0001$, $\eta^2=0.50$), the peak PF moment
177 ($P<0.0001$, $\eta^2=0.42$); and for the overall sagittal plane impulse ($P<0.0001$, $\eta^2=0.37$). In the
178 frontal plane, there were significant differences in the peak ADD moment ($P<0.0001$,
179 $\eta^2=0.30$) and time ($P=0.007$, $\eta^2=0.13$); and for the overall frontal plane impulse ($P<0.0001$,
180 $\eta^2=0.34$). In the transverse plane, there were significant differences in the peak EXT moment
181 ($P=0.019$, $\eta^2=0.14$) and time ($P<0.0001$, $\eta^2=0.35$); and for the overall transverse plane
182 impulse ($P=0.014$, $\eta^2=0.15$). Additionally, the total negative ($P<0.0001$, $\eta^2=0.61$) and
183 positive ($P=0.002$, $\eta^2=0.20$) work performed about the ankle joint was also significantly
184 different between conditions.

185 At the knee, there were significant differences in the peak KE moment ($P<0.0001$,
186 $\eta^2=0.28$) and time ($P=0.001$, $\eta^2=0.28$); and the peak KF moment ($P<0.0001$, $\eta^2=0.30$). At the
187 hip, there were significant differences in the peak HE moment ($P=0.001$, $\eta^2=0.24$), the peak
188 HF moment ($P<0.0001$, $\eta^2=0.41$) and time ($P=0.006$, $\eta^2=0.18$); and for the overall sagittal
189 plane impulse ($P=0.025$, $\eta^2=0.13$).

190

191 3.3 Centre of Pressure

192 A condition effect was found for A-P COP trajectory during early stance ($P<0.0001$,
193 $\eta^2=0.39$). Specifically, in the FF condition the COP was significantly anterior compared to
194 both KAL ($P<0.0001$) and BIRK ($P<0.0001$) (Fig. 4). No differences were evident between
195 conditions during mid-stance ($P>0.05$). In late stance a condition effect was again noted
196 ($P=0.025$, $\eta^2=0.16$), but the difference reached significance only between KAL and BIRK
197 conditions ($P<0.0001$).

198 Similarly, condition effects were also evident during early ($P=0.033$, $\eta^2=0.12$) and
199 late stance ($P<0.0001$, $\eta^2=0.54$) in the M-L direction. Specifically, FF COP was significantly
200 lateral than BIRK COP ($P=0.013$) during early stance, and significantly medial to both KAL
201 ($P=0.023$) and BIRK ($P<0.0001$) COP during late stance (Fig. 4).

202

203 4. Discussion

204 The purpose of the present study was to evaluate the biomechanical characteristics of
205 gait whilst walking in FitFlop™ footwear (FF). Comparisons were made against a
206 conventional sandal (BIRK) and a standard athletic trainer (KAL). Amongst the numerous
207 significant pair-wise differences reported in this study, there were four main findings. When
208 compared to both BIRK and KAL conditions, FF interaction results in: 1) a greater anterior
209 displacement of the average COP trajectory during early stance; 2) a greater medial
210 displacement of the average COP trajectory during late stance; 3) an increased peak hip
211 extensor moment during early stance; and 4) a reduction in sagittal plane ankle joint range of
212 motion throughout stance phase. In regard to (4) there is a corresponding reduction in torque

213 about the ankle joint and in the total negative work performed at this joint throughout stance
214 phase. Combined, these main findings indicate that FFs alter an individual's gait pattern
215 significantly, which corroborates our experimental hypothesis.

216 Microwobbleboard™ technology, the central technology underpinning FitFlop™
217 footwear, was designed to project the wearer into the middle section of the foot-bed (the
218 softer, “questioning” zone) earlier in stance phase than conventional footwear. The anterior
219 displacement of the COP trajectory during early stance, compared to both KAL and BIRK,
220 demonstrates that this objective has been successfully translated into a feature of gait whilst
221 wearing FF. This result, however, is in disagreement with the findings of Price et al. (2013)
222 who reported no shift in A-P COP trajectory for FF wearers. Instability in FitFlop™ footwear
223 was designed to be phase-dependent within the stance phase rather than across the whole
224 period of stance. Price and colleagues do not take account of this. Indeed, in the present study
225 A-P COP trajectory was shown to be different between conditions during early but not during
226 mid-stance. This means that forward progression was impeded during the transition between
227 early and mid-stance in the FF condition. Most likely this is caused by the softer middle
228 section of the foot-bed in FitFlop™ footwear. Expressing the COP trajectory relative to the
229 entire stance duration does not provide a sufficiently detailed representation of FF-interaction
230 dynamics, and important features of interaction may well be masked by such an approach.

231 The location of COP under the foot is a direct reflection of the neural control of the
232 ankle muscles (Winter, 1995). The more anteriorly directed average COP trajectory observed
233 in FF during early stance likely results from a reduced internal ankle dorsiflexion moment for
234 this footwear (Table 2). The magnitude of this moment determines the quantity of trunk
235 energy to be re-distributed to distal segments for effective deceleration during loading
236 response (Siegel et al., 2004). Hence this finding implies that for FF the second ankle rocker

237 is achieved earlier and a greater time is spent in single limb support. The unaltered average
238 A-P COP trajectory during mid-stance and the comparable stance phase durations between
239 conditions indirectly support this view. However, a reduced internal dorsiflexor moment
240 during loading response is an inherent characteristic of open-heel footwear designs (Zhang et
241 al., 2013). The present results show this is not the case whilst wearing a BIRK sandal. We
242 believe that the rigid construction of the BIRK sandal impedes forward progression during
243 early stance. In contrast to FF, a greater frontal plane internal adductor moment acting about
244 the ankle joint was evident, which likely reflects a control exerted by BIRK for ‘over-
245 pronation’. If this is true, then enhanced contribution from the plantarflexor muscles is
246 required to accelerate the body’s centre of mass (Wang and Gutierrez-Farewik, 2011). Whilst
247 this cannot be supported directly, we did observe significantly higher sagittal plane angular
248 impulse and total positive work performed about the ankle joint in the BIRK condition (Table
249 2). Moreover, the peak internal knee extensor moment, at approximately 20% of stance
250 phase, was also higher. The magnitude of this moment is negatively correlated with A-P COP
251 forward displacement ($r=-0.62$; $P=0.006$) (Shimokochi et al., 2009). This suggests
252 inefficiency in weight transfer from early stance to mid-stance whilst wearing a conventional
253 sandal compared to FitFlop[™] footwear.

254 The medio-lateral (M-L) destabilisation resulting from interaction with the FitFlop[™]
255 mid-sole construction is expected to result in enhanced activity of stabilising leg musculature.
256 In the present study, this consideration is investigated by way of the observed changes in the
257 net joint moments to account for the contribution of all muscles that act about a lower
258 extremity joint. We opted for this approach rather than using EMG since the smaller deeply
259 located muscles which may contribute to the control for a footwear-induced perturbation are
260 inaccessible with surface EMG electrodes. Whilst increased muscle activation cannot be
261 specifically revealed from the present findings, the joint moment data (limitations accepted)

262 provide evidence of greater reliance on the hip extensors for support and stabilisation when
263 wearing FFs than for comparative footwear. A greater internal hip extensor moment was
264 observed during early stance in the FF condition compared to both BIRK and KAL
265 conditions. The time at which this occurred corresponded to approximately 10% of stance
266 phase, which is the period when COP in the FF condition was found to be more anteriorly
267 displaced. This may explain the subsequent impediment to A-P COP displacement in mid-
268 stance as a means of preserving the overall support moment (Winter, 1980). Interestingly,
269 from the present study, the amplitudes of peak hip extensor moments appear inversely related
270 to the amplitudes of the subsequent peak knee extensor moments (Table 2, all conditions).
271 Shimokochi et al. (2009) have demonstrated such a relationship ($r=-0.66$, $P=0.003$) for the
272 support phase of a single limb landing task. Hence, it appears that wearing FitFlop™ footwear
273 alters lower extremity joint contributions: FF favours a support moment strategy for the hip,
274 whereas BIRK and KAL favour support moments for the knee.

275 Evidence published so far cannot confirm increased activation of the larger muscles of
276 the lower extremity whilst walking in FitFlop™ footwear (Burgess and Swinton, 2012; Price
277 et al., 2013). The results of Burgess and Swinton (2012) should be interpreted with caution,
278 since placing a restriction on participant walking speed (1.34 m/s) does not appear to be
279 ecologically valid given natural variation between subjects. Nonetheless, in their discussion
280 Burgess and Swinton (2012) allude to the potential for increased activation of the smaller,
281 deeper-lying muscles that act about the ankle joint whilst wearing FitFlop. Similar opinion
282 has been expressed in the literature with respect to instability shoes more generally (Nigg et
283 al., 2012). These muscles acting across the ankle joint complex react more quickly to frontal
284 and transverse plane changes in joint position than the larger muscles that ostensibly control
285 for sagittal plane deviations (Nigg et al., 2012). Joint stability is achieved with low levels of
286 torque, since these muscles have smaller moment arms. The reduction in peak internal ankle

287 plantarflexor moment observed in the FF condition and the subsequent kinematic adaptations
288 are potentially a consequence of a greater reliance placed on the smaller muscles crossing the
289 ankle joint. Until EMG investigations using indwelling electrodes or highly-selective surface
290 EMG array techniques (Coza et al., 2009) are available for dynamics studies this will remain
291 conjecture.

292 The most notable alteration to gait pattern observed with FitFlop™ in the present
293 study, was a reduction in peak ankle dorsiflexion angle (Table 1). The effect size for our
294 cohort was 76%. Whilst the participants in this study had ‘normal’ gait, the clinical
295 implications of this finding are worthy of mention. For example, reduced ankle dorsiflexor
296 range of motion is an important risk factor for individuals suffering from plantar fasciitis, the
297 most common foot-related disorder treated by healthcare professionals (McPoil et al., 2008).
298 Reducing sagittal plane ankle joint ROM and the corresponding net rotational peak force
299 acting about this joint during stance phase may be an effective method of reducing pain for
300 these sufferers. Price et al. (2013), however, found no differences in sagittal plane ankle joint
301 kinematics in their comparison of FitFlop™ and alternative instability footwear. . It is
302 noteworthy that these authors used a static neutral configuration prior to dynamic trials for
303 each condition. The present study, in contrast, performed this calibration only once during
304 barefoot standing, representing a global neutral configuration. This may explain the lack of
305 coherence in significant kinematic findings between the respective two studies.

306 Finally, we observed a significantly more medial COP trajectory during late stance in
307 the FF condition than in both KAL and BIRK. This matches expectations since all supinatory
308 rotations about the ankle joint (dorsiflexion, adduction and internal rotation) were
309 significantly restricted during the latter part of stance for participants wearing the FitFlop™
310 sandal. It is noteworthy that there was no difference in peak adduction between FF and

311 BIRK, but there was significantly greater transverse plane ankle joint motion in BIRK. The
312 latter is likely a compensation strategy imposed by the inherently stiff construction of the
313 BIRK sandal and this offsets the lack of frontal plane motion related to the foot-bed. The
314 present M-L COP data indicate that a greater range of motion was present in the FF condition
315 (Fig. 4). Similar findings have been presented by others for unstable footwear (Stoggl et al.,
316 2010; Zhang et al., 2012). If soft mid-sole constructions like that for FitFlop™ impair M-L
317 balance control (Perry et al., 2007), a mechanical balance control response would be expected
318 to allow unhindered forward progression. Indeed, Price et al. (2013) were able to demonstrate
319 a significantly greater m.peroneus longus (PL) activity during pre-swing in the FitFlop™
320 condition compared to other instability shoes. They did not, however, record any concomitant
321 differences between conditions in M-L COP range of motion. Unfortunately, the only other
322 FitFlop™ investigation Burgess and Swinton (2012) excluded PL activation from their
323 analysis and they reported no differences in activation profiles of the larger lower extremity
324 muscles. Future studies incorporating advanced EMG analysis may help to understand the
325 precise muscular responses to the perturbation induced by FitFlop™ footwear.

326 This study is not without limitations. The inverse dynamics procedure used to
327 calculate the net joint moments has a number of shortcomings, which can arise from errors in
328 basic methodological experimental procedures. For example, inaccuracies in ground reaction
329 force measurements and estimation of centre of pressure location; marker positioning,
330 selection of an appropriate technical frame of reference and skin movement artefact are all
331 significant contributors to the uncertainty in joint rotational force estimates derived through
332 this process (Riemer et al., 2008). Validation of the accuracy of the force platforms in 3-D
333 space, as performed in our Laboratory, overcomes the main sources of error relating to joint
334 moment calculation. Furthermore, all experimental conditions were performed by each
335 participant on the same day, which minimizes the potential error due to marker placement.

336 Finally, the errors associated with joint centre estimation and segmental motion tracking
337 during dynamic trials, were considered by adopting the CAST technique (Cappozzo et al.,
338 1995; Collins et al., 2009). Such steps remove the major experimentally-induced limitations
339 associated with the inverse dynamics procedure and give confidence in the reliability of the
340 study outcomes.

341

342 **5. Conclusion**

343 The present study has demonstrated that FitFlop™ footwear significantly alters gait
344 pattern for the wearers. The primary biomechanical response to the destabilising effect of the
345 mid-sole Microwobbleboard™ technology was the anterior displacement of the centre of
346 pressure trajectory. Stability, in preparation for mid-stance, appears to be consolidated
347 through larger net sagittal plane rotational forces about the hip. Consequently, the ankle joint
348 range of motion, the magnitude of peak dorsiflexion and the net rotational forces acting about
349 the ankle are reduced. This lowers the amount of work performed at the ankle joint during
350 support and propulsion.. These findings warrant future work to determine the potential
351 clinical benefits from reducing the ankle joint loading associated with walking in FitFlop™
352 footwear.

353

354 **Acknowledgement**

355 The authors would like to thank Dr. John Seeley for his assistance during the revision
356 process of this manuscript.

357

358 **Conflict of Interest**

359 Drs. Cook and James developed the Microwobbleboard™ technology concept in 2006
360 and Dr Cook is the named researcher on the patent, but without ownership of the intellectual
361 property. Between 2006-2013 London South Bank University and FitFlop Ltd. had a
362 contractual agreement in place for testing of FitFlop™ footwear. Drs Cook, James, or
363 London South Bank University receive no royalties for the commercial success of the
364 product. The present study was designed, analysed and the manuscript written by the authors
365 with no influence from the FitFlop Ltd.

366

367 **References**

368

369 Branthwaite, H., Chockalingam, N., Pandyan, A., Khatri, G., 2013. Evaluation of lower limb
370 electromyographic activity when using unstable shoes for the first time: a pilot quasi
371 control trial. *Prosthet. Orthot. Int.* 37, 275-281.

372 Burgess, K.E., Swinton, P.A., 2012. Do Fitflops increase lower limb muscle activity? *Clin.*
373 *Biomech.* 27, 1078-1082.

374 Cappozzo, A., Catani, F., Croce, U.D., Leardini, A., 1995. Position and orientation in space
375 of bones during movement: anatomical frame definition and determination. *Clin.*
376 *Biomech.* 10, 171-178.

377 Chang, R., Van Emmerik, R., Hamill, J., 2008. Quantifying rearfoot-forefoot coordination in
378 human walking. *J. Biomech.* 41, 3101-3105.

379 Collins, T.D., Ghoussayni, S.N., Ewins, D.J., Kent, J.A., 2009. A six degrees-of-freedom
380 marker set for gait analysis: repeatability and comparison with a modified Helen Hayes
381 set. *Gait Posture*. 30, 173-180.

382 Couillandre, A., Breniere, Y., 2003. How does the heel-off posture modify gait initiation
383 parameter programming? *J. Mot. Behav.* 35, 221-227.

384 Coza,A.,von Tscharnner,V.,Nigg,B.M., 2009. Activity mapping of lower leg muscles using a
385 circumferential electrode array. *Footwear Sci.* 1, 135-143.

386 Kamen, G., Caldwell, G.E., 1996. Physiology and interpretation of the electromyogram. *J.*
387 *Clin. Neurophysiol.* 13, 366-384.

388 Landry, S.C., Nigg, B.M., Tecante, K.E., 2010. Standing in an unstable shoe increases
389 postural sway and muscle activity of selected smaller extrinsic foot muscles. *Gait*
390 *Posture*. 32, 215-219.

391 Leardini, A., Benedetti, M.G., Berti, L., Bettinelli, D., Nativo, R., Giannini, S., 2007. Rear-
392 foot, mid-foot and fore-foot motion during the stance phase of gait. *Gait Posture*. 25,
393 453-462.

394 Lloyd, D.G., Besier, T.F., 2003. An EMG-driven musculoskeletal model to estimate muscle
395 forces and knee joint moments in vivo. *J. Biomech.* 36, 765-776.

396 Maffiuletti, N.A., 2012. Increased lower limb muscle activity induced by wearing MBT
397 shoes: physiological benefits and potential concerns. *Footwear Sci.* 4, 123-129.

398 McPoil, T.G., Martin, R.L., Cornwall, M.W., Wukich, D.K., Irrgang, J.J., Godges, J.J., 2008.
399 Heel pain--plantar fasciitis: clinical practice guidelines linked to the international

400 classification of function, disability, and health from the orthopaedic section of the
401 American Physical Therapy Association. *J. Orthop. Sports Phys. Ther.* 38, A1-A18.

402 Nigg, B., Federolf, P.A., von Tscharnner, V., Nigg, S., 2012. Unstable shoes: functional
403 concepts and scientific evidence. *Footwear Sci.* 4, 73-82.

404 Nigg, B., Hintzen, S., Ferber, R., 2006. Effect of an unstable shoe construction on lower
405 extremity gait characteristics. *Clin. Biomech.* 21, 82-88.

406 O'Connell, P.G., Lohmann Siegel, K., Kepple, T.M., Stanhope, S.J., Gerber, L.H., 1998.
407 Forefoot deformity, pain, and mobility in rheumatoid and nonarthritic subjects. *J.*
408 *Rheumatol.* 25, 1681-1686.

409 Perry, J., Burnfield, J.M., 2010. *Gait analysis: normal and pathological function*, 2nd ed.
410 SLACK, Thorofare, NJ.

411 Perry, S.D., Radtke, A., Goodwin, C.R., 2007. Influence of footwear midsole material
412 hardness on dynamic balance control during unexpected gait termination. *Gait Posture.*
413 25, 94-98.

414 Price, C., Smith, L., Graham-Smith, P., Jones, R., 2013. The effect of unstable sandals on
415 instability in gait in healthy female subjects. *Gait Posture.* 38, 410-415.

416 Riemer, R., Hsiao-Wecksler, E.T., Zhang, X., 2008. Uncertainties in inverse dynamics
417 solutions: a comprehensive analysis and an application to gait. *Gait Posture.* 27, 578-
418 588.

419 Romkes, J., Rudmann, C., Brunner, R., 2006. Changes in gait and EMG when walking with
420 the Masai Barefoot Technique. *Clin. Biomech.* 21, 75-81.

- 421 Seeley, M.K., Umberger, B.R., Shapiro, R., 2008. A test of the functional asymmetry
422 hypothesis in walking. *Gait Posture*. 28, 24-28.
- 423 Shimokochi, Y., Yong Lee, S., Shultz, S.J., Schmitz, R.J., 2009. The relationships among
424 sagittal-plane lower extremity moments: implications for landing strategy in anterior
425 cruciate ligament injury prevention. *J. Athl Train*. 44, 33-38.
- 426 Siegel, K.L., Kepple, T.M., Stanhope, S.J., 2004. Joint moment control of mechanical energy
427 flow during normal gait. *Gait Posture*. 19, 69-75.
- 428 Stefanyshyn, D.J., Stergiou, P., Lun, V.M., Meeuwisse, W.H., Worobets, J.T., 2006. Knee
429 angular impulse as a predictor of patellofemoral pain in runners. *Am. J. Sports Med*. 34,
430 1844-1851.
- 431 Stoggl, T., Haudum, A., Birklbauer, J., Murrer, M., Muller, E., 2010. Short and long term
432 adaptation of variability during walking using unstable (Mbt) shoes. *Clin. Biomech*. 25,
433 816-822.
- 434 Taube, W., Gruber, M., Gollhofer, A., 2008. Spinal and supraspinal adaptations associated
435 with balance training and their functional relevance. *Acta Physiol. (Oxf)*. 193, 101-116.
- 436 Wang, R., Gutierrez-Farewik, E.M., 2011. The effect of subtalar inversion/eversion on the
437 dynamic function of the tibialis anterior, soleus, and gastrocnemius during the stance
438 phase of gait. *Gait Posture*. 34, 29-35.
- 439 Winter, D.A., 1995. Human balance and posture control during standing and walking. *Gait &*
440 *Posture*. 3, 193-214.
- 441 Winter, D.A., 1980. Overall principle of lower limb support during stance phase of gait. *J.*
442 *Biomech*. 13, 923-927.

443 Zhang, S., Paquette, M.R., Milner, C.E., Westlake, C., Boyd, E., Baumgartner, L., 2012. An
444 unstable rocker-bottom shoe alters lower extremity biomechanics during level walking.
445 Footwear Sci. 4, 243-253.

446 Zhang, X., Paquette, M.R., Zhang, S., 2013. A comparison of gait biomechanics of flip-flops,
447 sandals, barefoot and shoes. J. Foot Ankle Res. 6, 45-1146-6-45.

448

449

450 **Table 1.** Mean (SD) spatio-temporal and ankle, knee and hip joint kinematic variables
451 measured during walking in an athletic shoe (KAL); FitFlopTM sandal (FF) and Birkenstock[®]
452 Gizeh sandal (BIRK).

	KAL	BIRK	FF	pair-wise comparisons
Stance time (s)	0.63 (0.05)	0.62 (0.04)	0.63 (0.04)	
Step length (m)	0.75 (0.06)	0.74 (0.06)	0.74 (0.06)	
Ankle				
<i>Sagittal</i>				
Peak Plantarflexion (°)	-8.0 (2.9)	-4.3 (2.6)	-7.3 (2.1)	FF > BIRK***, KAL > BIRK***
Peak Dorsiflexion (°)	10.1 (2.6)	11.8 (2.7)	7.8 (2.8)	FF < KAL***, FF < BIRK***, KAL < BIRK***
ROM PF-DF (°)	18.1 (3.1)	16.1 (2.5)	15.1 (2.7)	FF < KAL***, KAL > BIRK***
<i>Frontal</i>				
Peak Abduction (°)	-2.6 (2.5)	-3.3 (2.9)	-3.6 (2.9)	
Peak Adduction (°)	6.2 (4.2)	3.3 (3.6)	4.1 (3.7)	FF < KAL**, KAL > BIRK**
ROM ABD-ADD (°)	8.8 (3.4)	6.6 (3.4)	7.6 (3.2)	FF < KAL*, FF > BIRK**, KAL > BIRK***
<i>Transverse</i>				
Peak External Rotation (°)	-1.4 (9.4)	-2.1 (10.0)	-2.3 (9.8)	
Peak Internal Rotation (°)	14.4 (9.6)	15.4 (10.1)	13.7 (9.9)	FF < BIRK**
ROM EXT-INT (°)	15.8 (4.6)	17.5 (5.0)	16.0 (4.8)	FF < BIRK**, KAL < BIRK**
Knee				
Peak Flexion (°)	-16.9 (6.4)	-18.1 (6.0)	-16.8 (6.4)	FF < BIRK***, KAL < BIRK***
ROM IC-KF (°)	-14.3 (4.0)	-14.2 (3.5)	-13.2 (3.7)	FF < KAL**, FF < BIRK*
Hip				
Peak Extension (°)	-15.1 (3.4)	-15.2 (3.4)	-15.1 (3.2)	
ROM IC-HE (°)	-38.5 (5.0)	-38.6 (5.4)	-38.1 (5.3)	

453
454 * denotes $P < 0.05$, ** $P < 0.01$ and *** $P < 0.0001$.

455

456

457 **Table 2.** Mean (SD) ankle, knee and hip joint moment variables measured during walking in

458 an athletic shoe (KAL); FitFlopTM sandal (FF) and Birkenstock[®] Gizeh sandal (BIRK).

	KAL	BIRK	FF	pair-wise comparisons
Ankle				
<i>Sagittal</i>				
Peak DF Moment (Nm/kg)	0.200 (0.060)	0.203 (0.051)	0.177 (0.054)	FF < KAL*, FF < BIRK**
Time (% stance)	8.6 (1.1)	6.8 (1.3)	8.3 (1.5)	FF > BIRK***, KAL > BIRK***
Peak PF Moment (Nm/kg)	-1.414 (0.126)	-1.432 (0.120)	-1.348 (0.139)	FF < KAL*, FF < BIRK***
Time (% stance)	78.5 (1.7)	77.3 (1.7)	77.5 (3.5)	
Impulse (Nm/kg.s ⁻¹)	-0.316 (0.048)	-0.329 (0.045)	-0.304 (0.045)	FF < BIRK***, KAL < BIRK*
<i>Frontal</i>				
Peak ADD Moment (Nm/kg)	0.123 (0.031)	0.153 (0.047)	0.133 (0.039)	FF < BIRK*, KAL < BIRK***
Time (% stance)	39.1 (16.6)	33.7 (18.4)	28.4 (10.1)	FF < KAL**
Impulse (Nm/kg.s ⁻¹)	0.033 (0.018)	0.049 (0.022)	0.036 (0.021)	FF < BIRK**, KAL < BIRK***
<i>Transverse</i>				
Peak INT Moment (Nm/kg)	0.110 (0.039)	0.112 (0.041)	0.103 (0.032)	
Time (% stance)	11.4 (2.1)	10.6 (2.5)	11.5 (2.2)	
Peak EXT Moment (Nm/kg)	-0.660 (0.105)	-0.685 (0.117)	-0.649 (0.095)	FF < BIRK*
Time (% stance)	81.3 (2.0)	80.0 (1.7)	80.7 (1.7)	FF > BIRK*, KAL > BIRK***
Impulse (Nm/kg.s ⁻¹)	-0.127 (0.026)	-0.132 (0.025)	-0.122 (0.026)	FF < BIRK*
Total Negative Work (J.kg ⁻¹)	-0.168 (0.029)	-0.143 (0.021)	-0.128 (0.023)	FF < KAL***, FF < BIRK***, KAL > BIRK***
Total Positive Work (J.kg ⁻¹)	0.226 (0.041)	0.244 (0.048)	0.226 (0.043)	FF < BIRK*, KAL < BIRK**
Knee				
Peak KE Moment (Nm/kg)	0.433 (0.252)	0.475 (0.252)	0.422 (0.240)	FF < BIRK**, KAL < BIRK**
Time (% stance)	24.2 (2.5)	22.2 (1.6)	23.2 (3.8)	KAL > BIRK***
Peak KF Moment (Nm/Nm/kg)	-0.211 (0.118)	-0.236 (0.123)	-0.191 (0.111)	FF < BIRK**, KAL < BIRK**
Time (% stance)	63.2 (3.9)	62.5 (2.9)	60.1 (9.2)	
Impulse (Nm/kg.s ⁻¹)	0.051 (0.067)	0.049 (0.064)	0.050 (0.061)	
Hip				
Peak HE Moment (Nm/kg)	-1.215 (0.281)	-1.200 (0.278)	-1.270 (0.242)	FF > KAL*, FF > BIRK**
Time (% stance)	10.8 (2.4)	11.2 (2.4)	10.8 (2.3)	
Peak HF Moment (Nm/kg)	1.020 (0.213)	0.944 (0.195)	1.013 (0.187)	FF > BIRK***, KAL > BIRK***
Time (% stance)	86.7 (2.0)	85.6 (2.5)	86.3 (1.9)	KAL > BIRK*
Impulse (Nm/kg.s ⁻¹)	-0.027 (0.078)	-0.030 (0.072)	-0.037 (0.070)	FF > KAL*

459
460

* denotes $P < 0.05$, ** $P < 0.01$ and *** $P < 0.0001$.

461

462 **Figure 1. A:** Microwobbleboard™ technology is a triple density midsole engineered from
463 ethylene vinyl acetate (EVA) comprising a hard heel section (Shore A 45), a soft middle
464 section (Shore A 27) and an intermediate density at the toe region (Shore A 35). **B:** The three

465 conditions tested (left to right): a standard commercially-available athletic trainer (KAL), a
466 Birkenstock® Gizeh sandal (BIRK), and a FitFlop™ Walkstar sandal (FF). In KAL, windows
467 for sensor placement were cut at the 1st and 5th metatarsal head regions and calcaneus (not
468 visible).

469

470 **Figure 2.** Mean ($n=28$) joint angular kinematic profiles during stance phase. The shaded area
471 represents the standard deviation bandwidth of the athletic shoe (KAL); the FitFlop™ sandal
472 (FF) is denoted by the red line and the Birkenstock® Gizeh sandal (BIRK) by the thin black
473 line.

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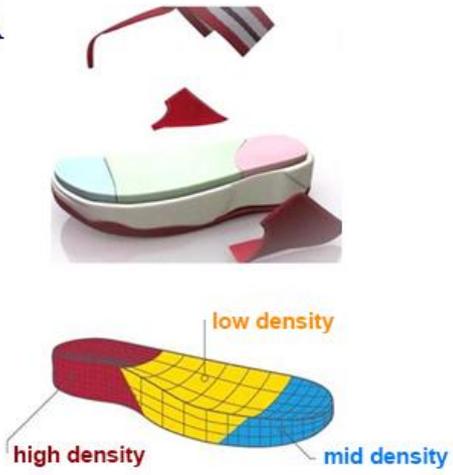
475 **Figure 3.** Mean ($n=28$) joint moment ensemble profiles during stance phase. The shaded area
476 represents the standard deviation bandwidth of the athletic shoe (KAL); the FitFlop™ sandal
477 (FF) is denoted by the red line and the Birkenstock® Gizeh sandal (BIRK) by then thin black
478 line.

479

480 **Figure 4.** Average anterior-posterior (A-P) and medio-lateral (M-L) centre of pressure (COP)
481 trajectories normalised (a.u) to foot length (proximal to distal joint centre) and width (distal
482 radius with respect to A-P axis), respectively. COP was expressed during early (0-33%), mid-
483 (34-66%) and late (67-100%) stance phase regions between conditions. * denotes $P<0.05$,
484 $**P<0.01$ and $***P<0.0001$.

485

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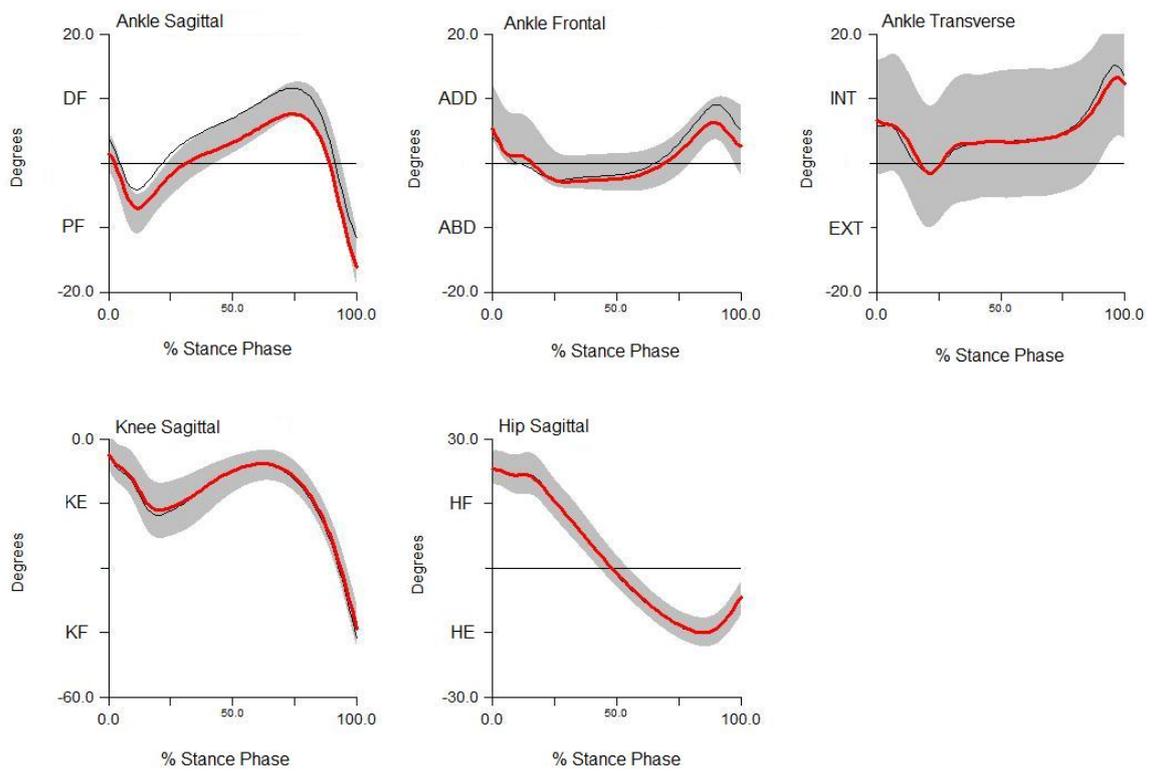


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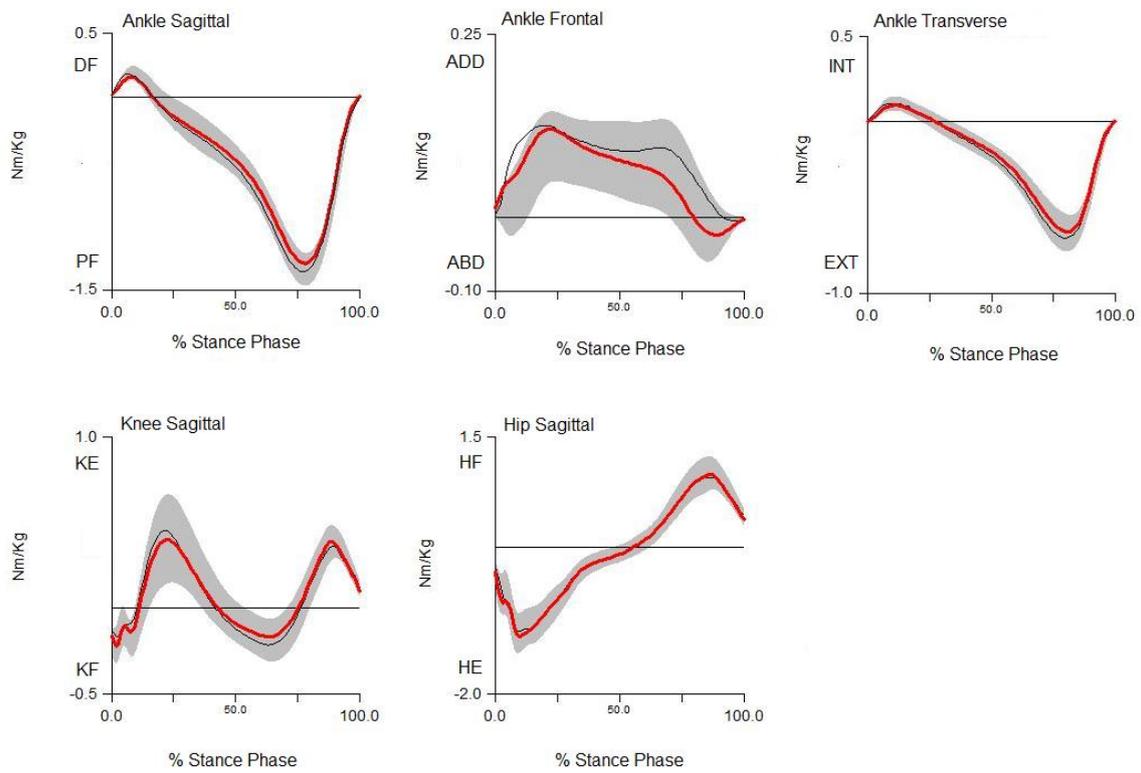
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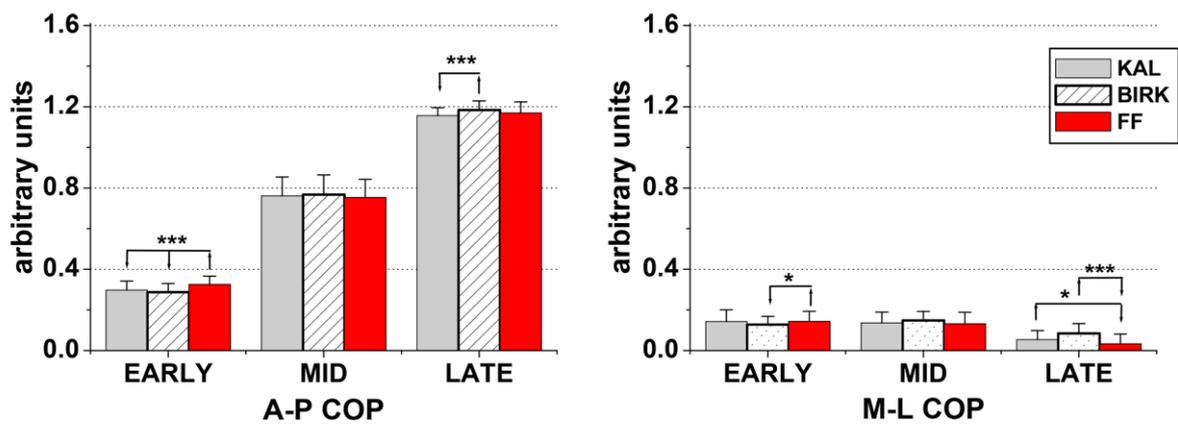
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Table 1

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	KAL	BIRK	FF	pair-wise comparisons
Stance time (s)	0.63 (0.05)	0.62 (0.04)	0.63 (0.04)	
Step length (m)	0.75 (0.06)	0.74 (0.06)	0.74 (0.06)	
Ankle				
<i>Sagittal</i>				
Peak Plantarflexion (°)	-8.0 (2.9)	-4.3 (2.6)	-7.3 (2.1)	FF > BRK ^{***} , KAL > BRK ^{***}
Peak Dorsiflexion (°)	10.1 (2.6)	11.8 (2.7)	7.8 (2.8)	FF < KAL ^{***} , FF < BRK ^{***} , KAL < BRK ^{***}
ROM PF-DF (°)	18.1 (3.1)	16.1 (2.5)	15.1 (2.7)	FF < KAL ^{***} , KAL > BRK ^{***}
<i>Frontal</i>				
Peak Abduction (°)	-2.6 (2.5)	-3.3 (2.9)	-3.6 (2.9)	
Peak Adduction (°)	6.2 (4.2)	3.3 (3.6)	4.1 (3.7)	FF < KAL ^{**} , KAL > BRK ^{**}
ROM ABD-ADD (°)	8.8 (3.4)	6.6 (3.4)	7.6 (3.2)	FF < KAL [*] , FF > BRK ^{**} , KAL > BRK ^{***}
<i>Transverse</i>				
Peak External Rotation (°)	-1.4 (9.4)	-2.1 (10.0)	-2.3 (9.8)	
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ROM IC-KF (°)	-14.3 (4.0)	-14.2 (3.5)	-13.2 (3.7)	FF < KAL ^{**} , FF < BRK [*]
Hip				
Peak Extension (°)	-15.1 (3.4)	-15.2 (3.4)	-15.1 (3.2)	
ROM IC-HE (°)	-38.5 (5.0)	-38.6 (5.4)	-38.1 (5.3)	

Table 2

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	KAL	BIRK	FF	pair-wise comparisons
Ankle				
<i>Sagittal</i>				
Peak DF Moment (Nm/kg)	0.200 (0.060)	0.203 (0.051)	0.177 (0.054)	FF = KAL [*] , FF = BRK ^{**}
Time (% stance)	8.6 (1.1)	6.8 (1.3)	8.3 (1.5)	FF = BRK ^{***} , KAL = BRK ^{***}
Peak PF Moment (Nm/kg)	-1.414 (0.126)	-1.432 (0.120)	-1.348 (0.139)	FF = KAL ^{**} , FF = BRK ^{***}
Time (% stance)	78.5 (1.7)	77.3 (1.7)	77.5 (3.5)	
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Peak ADD Moment (Nm/kg)	0.123 (0.031)	0.153 (0.047)	0.133 (0.039)	FF = BRK [*] , KAL = BRK ^{**}
Time (% stance)	39.1 (16.6)	33.7 (18.4)	28.4 (10.1)	FF = KAL ^{**}
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Total Positive Work (J.kg ⁻¹)	0.226 (0.041)	0.244 (0.048)	0.226 (0.043)	FF = BRK [*] , KAL = BRK ^{**}
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Peak KE Moment (Nm/kg)	0.433 (0.252)	0.475 (0.252)	0.422 (0.240)	FF = BRK ^{**} , KAL = BRK ^{**}
Time (% stance)	24.2 (2.5)	22.2 (1.6)	23.2 (3.8)	KAL = BRK ^{***}
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Time (% stance)	63.2 (3.9)	62.5 (2.9)	60.1 (9.2)	
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Peak HF Moment (Nm/kg)	1.020 (0.213)	0.944 (0.195)	1.013 (0.187)	FF = BRK ^{***} , KAL = BRK ^{***}
Time (% stance)	86.7 (2.0)	85.6 (2.5)	86.3 (1.9)	KAL = BRK [*]
Impulse (Nm/kg.s ⁻¹)	-0.027 (0.078)	-0.030 (0.072)	-0.037 (0.070)	FF = KAL [*]

Figure 1
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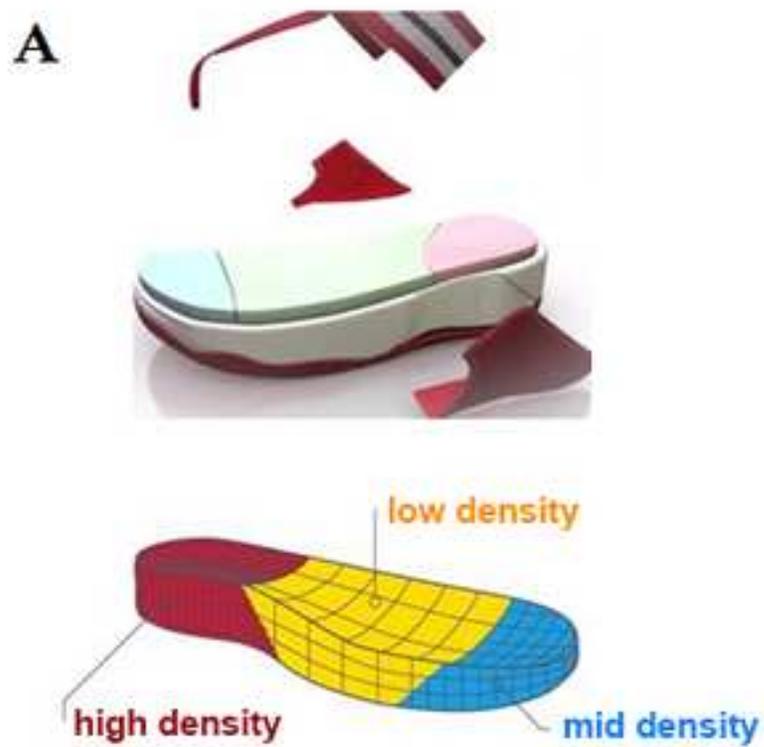


Figure 2
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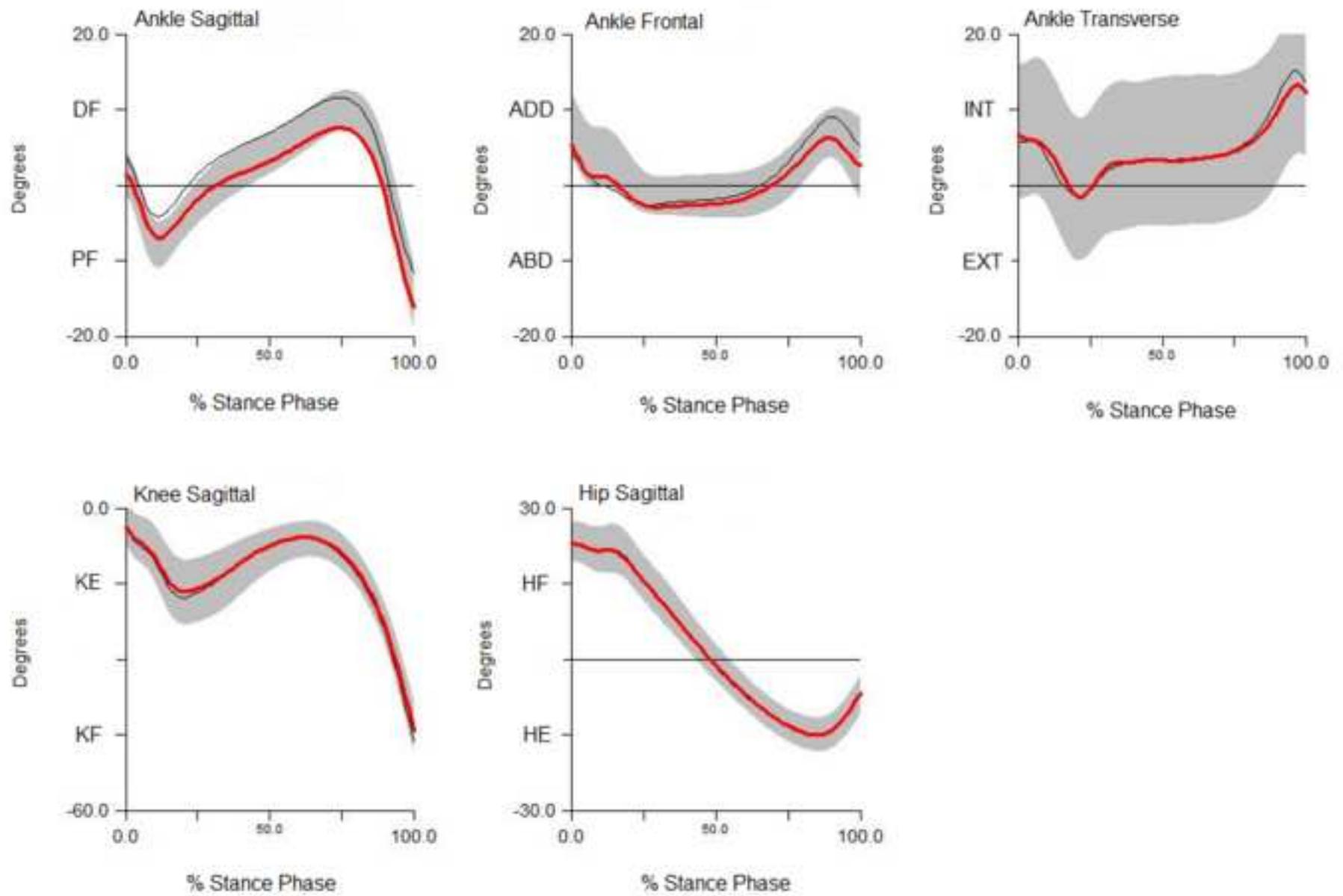


Figure 3
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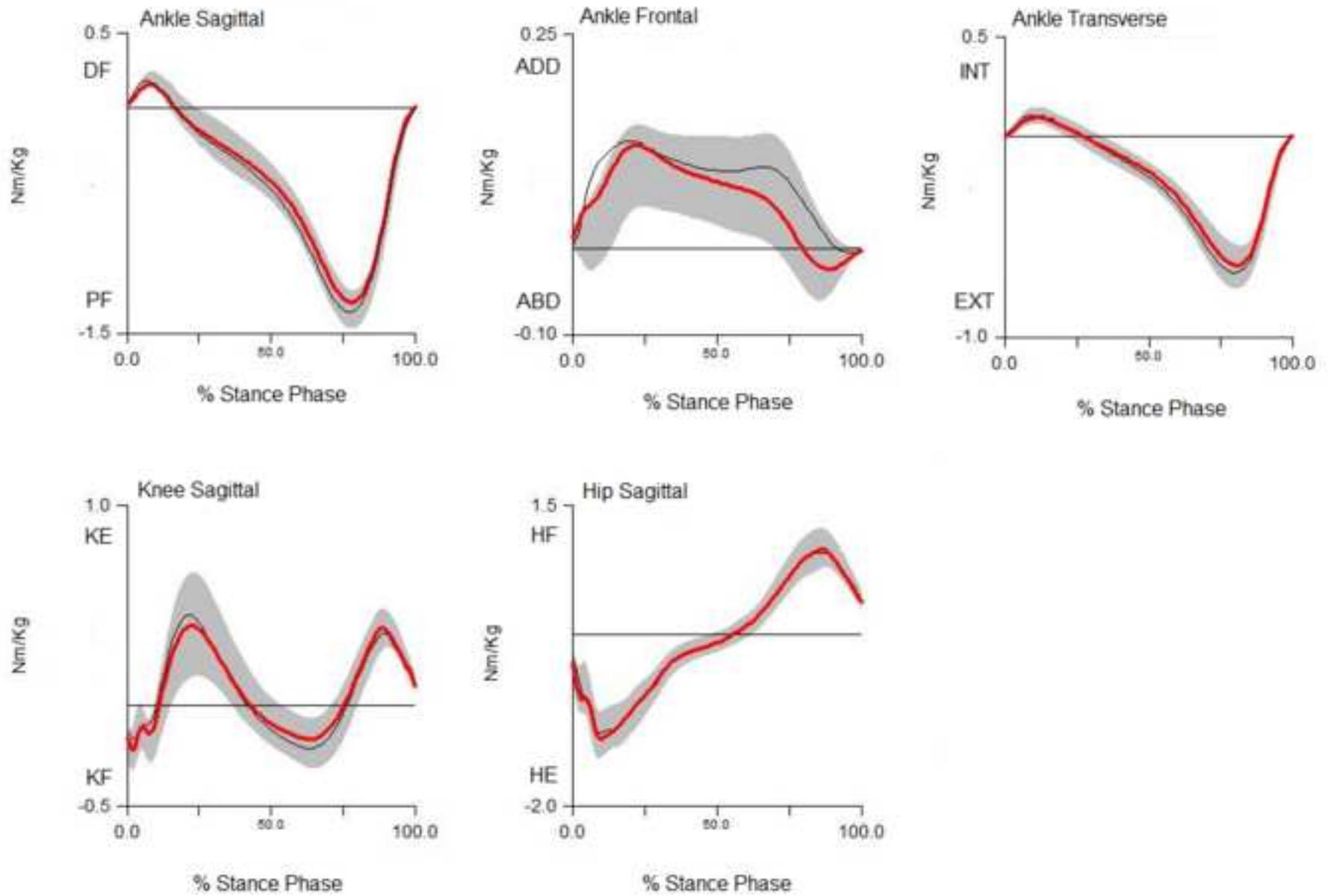


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