

Joint moments and power in the acceleration phase of bend sprinting

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1 **Title: Joint moments and power in the acceleration phase of bend sprinting**

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27 **Abstract**

28 Joint kinetics of the lower limb (hip, knee, ankle, midfoot and metatarsophalangeal joints)
29 were investigated during the acceleration phase of bend sprinting and straight-line sprinting.
30 Within the bend sprinting literature, it is generally accepted that sprint performance on the
31 bend is restricted by moments in the non-sagittal plane preventing the production of force in
32 the sagittal plane. However, there is limited evidence in conditions representative of elite
33 athletics performance that supports this hypothesis. Three-dimensional kinematic and ground
34 reaction force data were collected from seven participants during sprinting on the bend (36.5
35 m radius) and straight, allowing calculation of joint moment, power and energy. No changes
36 in extensor moment were observed at the hip and knee joints. Large effect sizes ($g = 1.07$)
37 suggest a trend towards an increase in left step peak ankle plantarflexion moment. This could
38 be due to a greater need for stabilisation of the ankle joint as a consequence of non-sagittal
39 plane adaptations of the lower limb. In addition, the observed increase in peak MTP joint
40 plantar-flexor moment might have implications for injury risk of the fifth metatarsal. Energy
41 generation, indicated by positive power, in the sagittal plane at the MTP and ankle joints was
42 moderately lower on the bend than straight, whilst increases in non-sagittal plane energy
43 absorption were observed at the ankle joint. Therefore, energy absorption at the foot and
44 ankle may be a key consideration in improving bend sprinting performance.

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52 **1. Introduction**

53 Research demonstrates a lower velocity during sprinting on the bend than straight
54 (Churchill et al., 2015; Churchill et al., 2016; Judson et al., 2019). At small radii (1-6 m),
55 the need to sustain muscle forces in the frontal and transverse planes is thought to prevent
56 sagittal plane moment generation and inhibit production of ground reaction force (Chang
57 and Kram, 2007). This hypothesis has not yet been confirmed under conditions
58 representative of competitive athletics, yet is generally accepted within the literature at
59 larger radii (Alt et al., 2015; Churchill et al., 2015).

60 Luo and Stefanyshyn (2012a) utilised wedged footwear, placing the left foot in a more
61 neutral position during sprinting (2.5 m radius), resulting in a lower eversion angle and
62 greater plantar-flexion moment than the control condition. Moreover, greater left limb non-
63 sagittal plane moments were observed during bend sprinting with a weighted vest condition
64 (Luo and Stefanyshyn, 2012b). Therefore, suggesting the left limb is able to generate more
65 force than observed during the control condition, but for some reason is prevented from
66 doing so. However, the radius evaluated (Luo and Stefanyshyn, 2012a) is smaller than those
67 during bend sprinting. Although Viellehner et al. (2016) reported greater left ankle plantar-
68 flexion moment with a 36.5 m radius compared to the straight, the submaximal effort is also
69 not representative of competitive performance. Therefore, a contradiction exists between
70 empirical evidence at smaller radii and submaximal velocity (Luo and Stefanyshyn, 2012a,
71 b; Viellehner et al., 2016), and hypotheses within the literature (Alt et al., 2015; Chang and
72 Kram, 2007; Churchill et al., 2015). Thus, analysis of 3D joint moments at maximal effort
73 and at radii representative of competitive athletics is required.

74 A proximal-distal sequencing of peak joint extension powers exists during the
75 acceleration phase of straight-line sprinting (Johnson and Buckley, 2001). High peak hip
76 adduction angles during bend sprinting (Alt et al., 2015; Churchill et al., 2015) could impact

77 the ability to produce forces in the sagittal plane and disrupt this proximal-distal
78 sequencing, resulting in a decrease in sprint performance. Furthermore, the complex
79 arrangement of the lower limb and associated non-sagittal plane adaptations could be risk
80 factors for injury during bend sprinting. Sprinters recorded the most muscular injuries in the
81 international athletics championships (2007 - 2015; Edouard, Branco and Alonso, 2016).
82 However, sprint events were recorded as a single category so bend specific injuries cannot
83 be identified. Moreover, iliotibial band syndrome and medial stress syndrome were amongst
84 frequent injuries in collegiate athletes undertaking training on an indoor track (Beukeboom,
85 Birmingham, Forwell and Ohrling, 2000). Analysis of joint kinetics would establish the net
86 demand of the joint and surrounding musculature, thus being influential in developing more
87 specific strength and conditioning and injury prevention programmes. Strength training for
88 sprint performance has focussed upon exercises in the sagittal plane which require triple
89 extension (Young, 2006). Resisted sled training is completed in a straight-line, and
90 plyometric exercises which do adopt a single leg approach make no suggestion of leg-
91 specific exercises (Young, 2006). This is similar to a further review (Wild et al., 2011),
92 suggesting elements of sprint training that comprise a change of direction or are limb
93 specific do not currently occur. Competitive bend sprinting takes place in an anti-clockwise
94 direction, meaning the left limb is always the inside limb, and continued sprinting in the
95 same direction is likely to result in limb-specific differences that warrant individual training
96 interventions. Therefore, this study aimed to investigate lower limb joint kinetics during
97 sprinting on the bend and straight. Conforming to the empirical evidence presented, it was
98 hypothesised that increased frontal and transverse plane moments would be greater on the
99 bend compared with the straight.

100 2. Methods

101 2.1. Participants

102 Following institutional ethical approval, seven male experienced bend sprinters (mean
103 age 22 ± 4 years; body mass 68.32 ± 6.98 kg; stature 1.79 ± 0.06 m; 200 m personal best
104 time: 21.8 - 23.43 s) participated in the study. Participants **had no history of injury within the**
105 **six months prior to data collection** and provided written informed consent.

106 **2.2. Experimental set-up**

107 Kinematic data were collected with a 15-camera optoelectronic system (13 x Raptor
108 and 2 x Eagle, Motion Analysis Corporation, CA, USA, 200 Hz). The direction of
109 progression was most closely aligned with the positive *x*-axis, the *y*-axis was vertical and the
110 *z*-axis mediolateral. The calibration volume (7 m long, 3 m wide and 1.5 m high) was located
111 tangentially to the apex of the curve to record data through the 10 – 17 m section of the 30 m
112 sprints (**Figure 1**). Torso, pelvis, thighs, shanks and feet segments (toebox, forefoot, rearfoot)
113 were modelled using a lower limb and trunk marker set (Judson et al., 2017). **Participants**
114 **were topless and wore form fitting shorts, allowing marker placement directly onto the skin.**
115 **Foot markers were placed onto the participants' preferred running spike and thought to**
116 **represent movement of the underlying bones.** Force data (1000 Hz) were collected with a
117 Kistler force plate (9287BA, 900 x 600 mm) embedded into the track surface at **11 - 13 m**
118 **from the start line.**

119 **2.3 Protocol**

120 For bend trials, a bend replicating lane 1 (radius 36.5 m) of a standard 400 m running
121 track (IAAF, 2008) was reconstructed on a flat section of indoor track. Simulating
122 **competitive bend sprinting, athletes completed bend trials in the anti-clockwise direction (left**
123 **turn).** Participants completed their typical competition warm-up before performing a
124 maximum of six (**three left steps and three right steps**) 30 m sprints at maximal effort in each
125 condition (bend and straight) **in a randomised order.** Starting blocks and an ‘*on your marks,*
126 *set, go*’ signal were used. One researcher modified the **location of the start blocks (to a**

127 maximum of 1 m forwards or backwards) to aid the athlete in contacting the force plate.

128 Approximately eight minutes were allowed between trials to avoid the onset of fatigue.

129 2.4 Data processing

130 Raw 3D marker coordinate data were analysed using Cortex software (version 5.3,
131 Motion Analysis Corporation, CA, USA). Visual 3D (version 6, C-Motion, MD, USA) was
132 used to define and construct segments, local coordinate systems and joint centres in line with
133 ISB guidelines (Wu et al., 2002; Wu et al., 2005). However, the multi-segment foot was
134 defined in accordance with Cappozzo et al. (1995).

135 Body segment parameters were estimated from de Leva (1996), estimates for a single
136 segment foot were applied to each segment of the multi-segment foot based upon individual
137 segment length (Deschamps et al., 2017; Dixon et al., 2012). Multi-segment foot values were
138 adjusted by 150 to 189 g (according to manufacturer specification) representing the mass of
139 individual participants' spiked shoes (Hunter et al., 2004a) and distributed based on the
140 relative segment length (Bezodis et al., 2012; Bruening et al., 2012). All data were filtered
141 with a low-pass, fourth order recursive Butterworth filter (18 Hz).

142 One successful trial per condition and per participant was analysed, as with previous
143 sprint research (Johnson and Buckley, 2001). In the current study, analysis of multiple trials
144 was not possible since participants did not always make contact with the force plate more
145 than once. A successful trial was defined as the participant making contact with the centre of
146 the force plate without the presence of targeting. Where multiple successful trials were
147 available, the first successful trial was used for analysis. *Touchdown* and *take-off* were
148 identified from vertical ground reaction force using a threshold mean plus two standard
149 deviations of data where there was zero load on the force plate (Bezodis et al., 2007) All
150 variables were calculated separately for the left and right step. All participants exhibited a
151 foot-strike where the toebox segment was the first point of contact.

152 *Joint moments* were calculated in Visual 3D (version 6, C-Motion, Rockville, MD,
153 USA) and expressed in the joint coordinate system, discussed in Schache and Baker (2007).
154 Data were cropped to the propulsive phase of stance and the entire ground reaction force
155 allocated to the toebox segment. Joint moment data were normalised to body weight and
156 height to maintain consistency with sprint literature (Charalambous et al., 2012).

157 *Joint powers* in each direction were calculated in Matlab (v2017a, Mathworks,
158 Natick, USA) as the dot product of non-normalised joint moment and joint angular velocity
159 and normalised using the following equation (Hof, 1996), adapted by Bezodis et al. (2010):

$$\frac{\vec{P}}{m \cdot g^{3/2} \cdot h^{1/2}}$$

160 where \vec{P} is power, m and h are mass and height of the sprinter and g is gravitational
161 acceleration.

162 **2.5 Statistical analysis**

163 Two-way repeated measures analysis of variance (ANOVAs) were performed where
164 condition (bend vs. straight) x limb (left vs. right) were analysed. Effect size (Hedges' g) was
165 used to indicate the magnitude of the effect, interpreted using $g < 0.20$ represents a trivial
166 difference, $0.20 \geq 0.50$ a small difference, $0.50 \geq 0.80$ a moderate difference and ≥ 0.80 a
167 large difference between means (Cohen, 1988).

168 **3. Results**

169 **Mean sprint velocities on the straight were 7.96 ± 0.23 m/s (left) and 8.00 ± 0.20 m/s**
170 **(right), and 7.81 ± 0.30 m/s (left), 7.89 ± 0.34 m/s (right) on the bend.** Joint moment and
171 power across the propulsive phase of stance are shown in Figures 1-4. For peak MTP joint
172 plantar-flexor moment, the condition x limb interaction was non-significant ($F_{(1, 6)} = 0.06$, p
173 $= 0.81$). There was a main effect for limb ($F_{(1, 6)} = 8.46$, $p = 0.03$, $g = 1.90$) due to a larger
174 plantar-flexor moment in the right MTP joint than left. Moderate and large effect sizes ($g =$

175 0.59, 1.45) suggest a trend towards a greater plantar-flexor moment on the bend than straight
176 in the right and left MTP joints, respectively. A large effect size ($g = 1.07$) suggests a trend
177 towards increased peak left step ankle plantar-flexion moment on the bend compared with the
178 straight, although the interaction was non-significant ($F_{(1,6)} = 2.33, p = 0.18$).

179 For peak ankle eversion moment, the condition x limb interaction was non-significant
180 ($F_{(1,6)} = 0.70, p = 0.43$). A main effect for limb was observed, $F_{(1,6)} = 26.00, p < 0.01$, due
181 to an increase in left step peak ankle eversion moment on the bend compared with the right
182 step on the bend ($g = 1.23$). A large effect size suggests a greater left step peak ankle eversion
183 moment on the bend than straight ($g = 0.82$). The condition x limb interaction for midfoot
184 eversion was non-significant ($F_{(1,6)} = 3.20, p = 0.12$). However, moderate and large increases
185 in left step peak midfoot eversion moment were observed during bend sprinting compared
186 with the straight ($g = 0.65$) and also when compared with the right step on the bend
187 ($g = 1.31$).

188 There was no condition x limb interaction ($F_{(1,6)} = 0.57, p = 0.48$) for peak hip flexor
189 moment. Although the main effect for condition was non-significant ($F_{(1,6)} = 4.47, p = 0.08$),
190 a large and moderate ($g = 1.10, 0.70$) decrease in peak hip flexor moment was observed for
191 the left and right steps on the bend compared with the straight, respectively. For peak knee
192 flexor moment a large decrease was observed during the left step of bend sprinting. However,
193 the condition x limb interaction was non-significant ($F_{(1,6)} = 4.66, p = 0.07$). For hip
194 adduction, the condition x limb interaction was non-significant ($F_{(1,6)} = 2.84, p = 0.14$).
195 However, there was an increase ($g = 0.85$) in peak left step hip adductor moment during
196 sprinting on the bend relative to the straight.

197 **3.1 Joint power**

198 At the hip joint, no condition x limb interactions for peak power in the sagittal plane
199 (positive: $F_{(1,6)} = 0.06, p = 0.82$, negative: $F_{(1,6)} = 0.23, p = 0.65$) were observed. Although

200 there was no condition x limb interaction ($F_{(1,6)}=1.59, p = 0.25$), there was a trend towards a
201 greater left step peak positive hip power in the frontal plane during ($g = 0.82$) on the bend
202 than straight (main effect, limb: $F_{(1,6)} = 6.29, p = 0.05$). A moderate ($g = 0.63$) effect size
203 suggests a trend towards greater left step peak negative hip power in the transverse plane
204 during bend sprinting relative to the straight, however no condition x limb interaction was
205 reported ($F_{(1,6)} = 2.88, p = 0.14$).

206 At the ankle, there was no condition x limb interaction for peak negative sagittal plane
207 joint power ($F_{(1,6)} = 0.05, p = 0.841$). There was a large, non-significant, increase in left step
208 sagittal plane ankle energy absorption on the bend compared with the straight (main effect
209 limb: $F_{(1,6)}=3.287, p = 0.13, g = 0.80$). The condition x limb interaction was non-significant
210 for peak positive transverse plane ankle power ($F_{(1,6)} = 1.13, p = 0.34$). However, there was
211 a moderate increase in peak positive left step ankle power in the transverse plane during bend
212 sprinting relative to the straight (main effect limb: $F_{(1,6)} = 4.800, p = 0.08, g = 0.76$).

213 At the MTP joint, no condition x limb interaction was found ($F_{(1,6)} = 0.261,$
214 $p = 0.627$). However, there were large and moderate ($g = 0.95, 0.52$) increases in peak
215 negative joint power for the left and right step, respectively (main effect condition:
216 $F_{(1,6)} = 6.27, p = 0.05$). There was an increase in MTP joint energy absorption (main effect
217 condition: $F_{(1,6)} = 7.14, p = 0.04$) on the bend compared with the straight (left, $g = 1.68,$
218 right, $g = 0.30$). For midfoot peak positive power, there was a main effect for condition, $F_{(1,6)}$
219 $= 48.04, p < 0.01$, with an increase in peak positive midfoot power in both the left ($g = 1.20$)
220 and right ($g = 0.74$) step on the bend compared with the straight. For peak negative midfoot
221 power, there was a main effect for limb ($F_{(1,6)} = 19.683, p < 0.01$, with an increase in peak
222 negative midfoot power in the left step compared with the straight ($g = 1.45$).

223 4. Discussion

224 This study evaluated joint kinetics during the acceleration phase of sprinting on the
225 bend and straight. During bend sprinting, there was a large, but non-significant, decrease in
226 peak flexor moment of the left hip and knee compared with straight-line sprinting. There was
227 a moderate decrease in peak flexor moment of the right hip, although no change was
228 observed at the right knee. Changes in non-sagittal plane moments were also observed, with a
229 trend towards an increase in peak left hip adduction moment.

230 During both bend and straight conditions, sagittal plane hip moment was extensor for
231 the majority of the propulsive phase of stance, becoming flexor towards the latter stages of
232 stance, similar to early (Bezodis et al., 2014), to mid-acceleration (Johnson and Buckley,
233 2001), and maximal speed (Bezodis et al., 2008) straight-line sprinting. A large effect size
234 suggests a lower left hip flexor moment during sprinting on the bend than straight. The hip
235 flexor moment towards the end of stance drives the limb forward during the swing phase
236 (Charalambous et al., 2012). Consequently, during bend sprinting athletes may experience
237 difficulties repositioning the left leg due to the kinematic alterations such as greater hip
238 adduction and external rotation observed in Judson et al. (2019). Therefore, swing phase
239 mechanics might increase understanding bend sprinting performance.

240 Moderate effect sizes suggest a trend towards a greater left step peak negative frontal
241 and transverse plane hip power on the bend than on the straight. This negative power
242 indicates an eccentric contraction of the muscles surrounding the hip, which could stabilise
243 the pelvis during sprinting on the bend (Segal et al., 2009). The pelvis is mutually influenced
244 by each limb, and the left and right limb have been shown to behave differently during bend
245 sprinting with the left limb characterised by a high peak hip adduction angle (Alt et al., 2015;
246 Churchill et al., 2015; Judson et al., 2019). Therefore, a greater level of pelvic control is
247 likely required to overcome these adaptations.

248 Ankle joint results suggest the limiting factor to sprint performance on the bend is
249 more complex than proposed by Chang and Kram (2007). A greater left step peak plantar-
250 flexor moment was observed on the bend than the straight, supporting results at submaximal
251 effort bend sprinting (Viellehner et al., 2016). These results dispute the idea that moment
252 production at the ankle joint is constrained by moments in the non-sagittal planes. Hunter et
253 al. (2004b) and Johnson and Buckley (2001) suggested ankle plantar-flexor moment acts
254 against anterior rotation of the shank to prevent the collapse of the shank due to the effect of
255 ground reaction force. The increase in left step peak ankle plantar-flexor moment is possibly
256 required to stabilise the shank as a consequence of the non-sagittal plane adaptations of the
257 lower limb. Furthermore, peak MTP plantar-flexor moment was greater during bend sprinting
258 than the straight. MTP moment was plantar-flexor for the duration of the propulsive phase of
259 stance, supporting research in the acceleration phase of straight-line sprinting (Smith et al.,
260 2014; Stefanyshyn and Nigg, 1997). MTP bending moments, attributed to high moments at
261 the forefoot (or toebox) during push-off, are greatest during acceleration movements and a
262 possible risk factor for fifth-metatarsal stress fractures (Orendurff et al., 2009). Bend
263 sprinting likely also increases the torsional load experienced by the long bones of the foot and
264 the increase in ankle and MTP plantar-flexor moments observed in the present study are a
265 risk factor for injury. Orendurff et al. (2009) recommendations for minimising injury risk
266 include careful consideration of the rate at which sprint training volume increases and the
267 number of accelerations performed within a session, and may have particular importance in
268 preparation for bend sprinting.

269 The ankle joint generated the most power in the sagittal plane (normalised power:
270 0.97 - 1.24) compared with the hip (0.52 - 0.56) and knee joints (0.09 - 0.12). These findings
271 agree with previous literature demonstrating the dominant role of the ankle joint in sprinting
272 (Brazil et al., 2017; Debaere et al., 2015; Dorn et al., 2012; Johnson and Buckley, 2001).

273 Large and moderate increases in left step peak positive power were observed at the ankle in
274 the frontal and transverse planes - this did not affect positive power production in the sagittal
275 plane where no differences were observed. Therefore, strengthening foot and ankle muscles
276 in non-sagittal planes specific to bend sprinting may be beneficial in improving bend
277 sprinting performance.

278 Compared with the hip and ankle joints, power generation at the knee joint was
279 relatively low, suggesting a supporting role at the knee and agree with Heinrich et al. (2015)
280 who proposed the knee may have a sub-unit function, facilitating the transfer of power from
281 the hip through to the ankle. Thus, similar to straight-line sprinting (Bezodis et al., 2008), the
282 knee functions in a supporting role during bend sprinting.

283 In both conditions, a proximal to distal sequence of peak extensor power generation
284 was observed. However, during bend sprinting with the left step, the timing of peak extensor
285 power at the knee was later than that of the straight. This sequential transfer of power is
286 thought to allow joints such as the ankle to achieve a higher power output despite having
287 relatively smaller muscles (Jacobs et al., 1996). Kinematic adaptations, such as increased left
288 step hip adduction (Alt et al., 2015; Churchill et al., 2015), might delay the transfer of power
289 from the hip to the knee. However, this increase in peak left hip frontal plane power is likely
290 a necessary consequence of the need to generate centripetal force and stay in the correct lane.

291 The negative sagittal plane power indicates large amounts of energy were absorbed in
292 the ankle and MTP joints in both the bend and straight conditions. Effect sizes suggest a trend
293 towards a greater negative power, and thus energy absorption, in the left MTP, midfoot and
294 ankle on the bend than the straight. During bend sprinting, the left step also demonstrated
295 greater energy absorption than the straight in the frontal plane at the midfoot and ankle, and
296 the transverse plane at the ankle. Therefore, energy absorption in the foot and ankle may be a
297 key consideration for improving bend performance. Smith et al. (2014) demonstrated

298 increased sprint velocity and energy generated at the MTP joint during push-off in a shod
299 condition compared with sprinting barefoot, suggesting sprint spikes are capable of
300 improving MTP joint function, resulting in faster sprint performance. Strength training
301 targeting muscles such as tibialis anterior and posterior, in addition to intrinsic foot muscles
302 such as abductor hallucis and flexor digitorum brevis, might provide a further opportunity to
303 influence sprint performance, particularly on the bend (Smith et al., 2014).

304 The small sample size of this study is comparable to previous literature (Alt et al.,
305 2015; Churchill et al., 2015; Churchill et al., 2016). Whilst an increased sample size would
306 be desirable in terms of statistical power, the inclusion criteria (200 m PB: 23.5 s) meant this
307 was not possible. Alt et al. (2015) imply velocity specific modulations are apparent during
308 bend sprinting. Consequently, extending the inclusion criteria to include less-skilled sprinters
309 may introduce variability into the sample and was not considered appropriate. **Bezodis et al.**
310 **(2019) suggested differences in sprint start kinematics were more indicative of differences in**
311 **performance levels, rather than sex. However, to enable the development of specific training**
312 **interventions, future work investigating the biomechanical adaptations of bend sprinting**
313 **specific to the female population is warranted.** Moreover, the analysis of multiple steps
314 throughout the acceleration phase would be preferable, **future research should consider using**
315 **alternative equipment to allow analysis of a greater number of steps across the acceleration**
316 **phase. In addition, increased longitudinal bending stiffness in sprint spikes has a localised**
317 **effect on the MTP joint (Smith et al., 2016). Therefore, different sprint spikes may introduce**
318 **between-participant differences. Although preferred sprint spikes are warranted to maintain**
319 **representativeness, future work might examine sprint spike stiffness specifically for bend**
320 **sprinting performance.**

321 In conclusion, this study demonstrates substantial changes to the function of the joints
322 of the lower limb and loading of the surrounding musculoskeletal structures during bend

323 sprinting compared with straight-line sprinting. Whilst peak flexor moments at the hip and
324 knee were lower during sprinting on the bend than straight, increased plantar-flexor moment
325 at the ankle and MTP suggest the limiting factor to sprint performance on the bend is a
326 complex interaction. Compared with straight-line sprinting, there was also an increase in non-
327 sagittal plane joint moments during bend sprinting, particularly at the hip and ankle joints. To
328 improve bend sprinting performance, athletes should consider developing the ability to
329 produce plantar-flexion from an internally rotated position specific to bend sprinting.

330

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440 **Conflict of interest statement**

441 The author's declare no conflict regarding the content of this article.

442

443 **Figure 1.** Plan view of the experimental set-up (not to scale).

444 **Figure 2.** Hip joint angle, joint moment and joint power for the left (red) and right (black) steps on the bend (dashed line - - -
445) and straight (solid line —). The shaded area indicates standard deviation; for clarity this is presented for the bend trials
446 only.

447 **Figure 3.** Knee joint angle, joint moment and joint power for the left (red) and right (black) steps on the bend (dashed line - - -
448 -) and straight (solid line —). The shaded area indicates standard deviation; for clarity this is presented for the bend trials
449 only.

450 **Figure 4.** Ankle joint angle, joint moment and joint power for the left (red) and right (black) steps on the bend (dashed line - - -
451 -) and straight (solid line —). The shaded area indicates standard deviation; for clarity this is presented for the bend trials
452 only.

453 **Figure 5.** Midfoot and MTP joint angle, joint moment and joint power for the left (red) and right (black) steps on the bend
454 (dashed line - - -) and straight (solid line —). The shaded area indicates standard deviation; for clarity this is presented for
455 the bend trials only.

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