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Title: Understanding the effect of touchdown distance and ankle joint kinematics on sprint acceleration performance through computer simulation

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1 **Abstract**

2 This study determined the effects of simulated technique manipulations on early acceleration
3 performance. A planar seven-segment angle-driven model was developed and quantitatively
4 evaluated based on the agreement of its output to empirical data from an international-level
5 male sprinter (100 m personal best = 10.28 s). The model was then applied to independently
6 assess the effects of manipulating touchdown distance (horizontal distance between the foot
7 and centre of mass) and range of ankle joint dorsiflexion during early stance on horizontal
8 external power production during stance. The model matched the empirical data with a mean
9 difference of 5.2%. When the foot was placed progressively further forward at touchdown,
10 horizontal power production continually reduced. When the foot was placed further back,
11 power production initially increased (a peak increase of 0.7% occurred at 0.02 m further
12 back) but decreased as the foot continued to touchdown further back. When the range of
13 dorsiflexion during early stance was reduced, exponential increases in performance were
14 observed. Increasing negative touchdown distance directs the ground reaction force more
15 horizontally; however, a limit to the associated performance benefit exists. Reducing
16 dorsiflexion, which required achievable increases in the peak ankle plantar flexor moment,
17 appears potentially beneficial for improving early acceleration performance.

18

19 200 words.

20

21 **Keywords:** kinematics, kinetics, modelling, sprinting, technique

22

23 **Introduction**

24 Sprinting is a pure athletic endeavour where overall performance is determined by the ability
25 to cover a short distance in the least possible time. The margins of success in international
26 sprinting are often small and technique adjustments which could result in slight performance
27 improvements are therefore of great interest to coaches and athletes. It has been demonstrated
28 that the production of maximum external power from the very first step of a sprint is the most
29 favourable strategy for optimum overall sprint performance (de Koning, de Groot, & van
30 Ingen Schenau, 1992; van Ingen Schenau, de Koning, & de Groot, 1994; van Ingen Schenau,
31 Jacobs, & de Koning, 1991). The techniques associated with a powerful early acceleration
32 phase are therefore of clear relevance to overall sprint performance.

33

34 Empirical research has recently supported the importance of technical ability for early
35 acceleration performance (Kugler & Janshen, 2010; Morin, Edouard, & Samozino, 2011;
36 Rabita et al., 2015). These studies identified that the ability to direct the resultant ground
37 reaction force (GRF) vector in a more horizontal direction was associated with higher levels
38 of sprint acceleration performance, whereas the magnitude of the GRF vector was not.
39 Furthermore, Kugler and Janshen (2010) suggested that a greater negative touchdown
40 distance, i.e. planting the stance foot more posterior relative to the centre of mass (CM) at
41 touchdown, facilitated a forward leaning position and the generation of greater horizontal
42 propulsive forces. The knee joint has been linked to forward lean during the first stance phase
43 (Debaere, Delecluse, Aerenhouts, Hagman, & Jonkers, 2013) and knee extensor moments and
44 powers have recently been identified as an important feature of the first stance phase in well-
45 trained and international-level sprinters (Bezodis, Salo, & Trewartha, 2014; Charalambous,
46 Irwin, Bezodis, & Kerwin, 2012; Debaere et al., 2013). It is therefore possible that any effects
47 of touchdown distance on early acceleration performance are related to knee joint mechanics

48 at touchdown. Whilst it has been proposed that a large negative touchdown distance is
49 favourable for early acceleration performance (Bezodis, Salo, & Trewartha, 2008; Kugler &
50 Janshen, 2010), this has only been based on descriptive differences between sprinters.
51 Furthermore, if such a strategy is beneficial for performance, it is conceivable that a limit
52 may exist and that sprinters should not continually strive to increase the negative touchdown
53 distance.

54

55 Whilst there are likely to be numerous aspects of technique which are important for early
56 acceleration performance, one other important aspect proposed in the literature relates to the
57 energy generating role of the ankle joint (Bezodis et al., 2014; Charalambous et al., 2012).
58 The ankle goes through dorsiflexion followed by plantar flexion whilst a net plantar flexor
59 moment is typically dominant throughout stance in every stance phase of a sprint. Bezodis et
60 al. (2014) identified that the ankle generates up to four times more energy than it absorbs
61 during the first stance phase compared to zero net energy generation during the mid-
62 acceleration phase (Johnson & Buckley, 2001) and net energy absorption during the
63 maximum velocity phase (Bezodis, Kerwin, & Salo, 2008). Charalambous et al. (2012) also
64 determined that a 'stiffer' ankle whilst dorsiflexing during the early part of the first stance
65 phase was positively related to higher horizontal CM velocities in a single sprinter. It is
66 therefore conceivable that a reduction in ankle joint dorsiflexion during early stance could be
67 another technical feature which is of benefit to early acceleration performance.

68

69 Whilst empirical evidence suggests that a large negative touchdown distance and reduced
70 ankle joint dorsiflexion during early stance may be beneficial for early acceleration sprint
71 performance, the effect of manipulating these aspects of technique on performance have not
72 been determined. This is likely due to an understandable resistance to exploratory

73 experimental manipulation of technique from coaches and athletes, particularly in elite sport
74 (Kearney, 1999). Simulation modelling offers an alternative approach which allows the
75 consideration of hypothetical situations to yield a more complete understanding than that
76 possible experimentally (Yeadon & Challis, 1994) and which provides valuable preliminary
77 evidence to determine the theoretical feasibility of potential applied interventions (Knudson,
78 Elliott, & Hamill, 2014). Such models are typically customised based on empirical data from
79 a single, highly-trained athlete. Model parameters and inputs are obtained from data on real
80 performances and quantitative evaluation of the model output is performed against relevant
81 aspects of the athlete's technique and performance. Angle-driven simulation models have
82 been previously used in other sports to systematically manipulate kinematic aspects of
83 technique and determine the consequent effects on performance (e.g. Gittoes, Brewin, &
84 Kerwin, 2006; Hiley & Yeadon, 2003a; 2003b; 2007). However, no such model exists which
85 has been specifically designed and evaluated to investigate sprinting. The primary aim of this
86 study was to determine the effects of manipulating touchdown distance and ankle joint
87 dorsiflexion during the first stance phase of a maximal effort acceleration out of blocks. It
88 was hypothesised that: 1) an increasingly large negative touchdown distance and 2) reduced
89 ankle joint dorsiflexion during early stance would each independently lead to increases in
90 first stance phase performance. In order to address this primary aim and these hypotheses, a
91 prerequisite aim was to develop a computer simulation model and evaluate its representation
92 of technique and performance during the first stance phase of a sprint.

93

94 **Methods**

95 Empirical data were collected from an international-level male sprinter (age, 20 years; mass,
96 86.9 kg; height, 1.78 m; 100 m personal best, 10.28 s) to obtain appropriate simulation model
97 parameters and allow a quantitative evaluation. The sprinter's ability to accelerate was

98 highlighted by the fact that he had reached the 60 m final of the European Indoor
99 Championships in the previous season. Three maximal effort 30 m sprints from blocks were
100 completed at an indoor track just prior to the competition phase of the indoor season. The
101 sprinter provided written informed consent to participate and the study was approved by the
102 University research ethics committee. Synchronised GRF and two-dimensional video data
103 were collected. A simulation model was developed using input data and parameters obtained
104 from the empirical collection and matching optimisations. Specific model outputs were
105 evaluated against empirical data to ensure that the model appropriately reflected reality.
106 Model-based simulations were then run by separately manipulating touchdown distance and
107 ankle joint dorsiflexion during early stance to determine their effects on first stance phase
108 performance and address the primary aim of this study.

109

110 ***Empirical data collection***

111 Two digital video cameras (MotionPro[®] HS-1, Redlake, San Diego, CA, USA) were
112 positioned in series with overlapping 2.5 m wide fields of view to obtain sagittal plane
113 images (1280 × 1024 pixels) from the ‘set’ position until the end of the first stance phase at
114 200 Hz. Ground reaction forces from the first stance phase were obtained at 1000 Hz using a
115 force platform (Kistler, 9287BA, 1000 Hz, Winterthur, Switzerland) covered with artificial
116 track surface. A third camera (MotionPro[®] HS-1, Redlake, San Diego, CA, USA) was
117 positioned perpendicular to the centre of the force platform to obtain images (800 × 600
118 pixels) of the lower leg and foot during ground contact inside a 0.9 m field of view. The three
119 video cameras and the force platform were synchronised to within 1 ms using a custom
120 designed trigger system. An experienced starter administered standard ‘*on your marks*’ and
121 ‘*set*’ commands before pressing a trigger button which sent a signal to initiate and
122 synchronise all devices and an audio signal to the athlete.

123

124 The raw video files were imported into digitising software (Peak Motus[®], v. 8.5, Vicon,
125 Oxford, UK), and were manually digitised at full resolution with a zoom factor of 2.0. For the
126 200 Hz cameras, twenty anatomical points were digitised (vertex, C7, shoulder, elbow, wrist,
127 hip, knee, ankle and metatarsal-phalangeal (MTP) joint centres, fingertips and toes) from 10
128 frames prior to movement onset until 10 frames after first stance toe-off. For the 1000 Hz
129 camera, the 5th MTP joint centre and toe were digitised from 10 frames prior to touchdown
130 until 10 frames after toe-off. The raw digitised co-ordinates were projectively scaled and the
131 resulting raw displacement time-histories were exported from Peak Motus[®] for subsequent
132 analysis in Matlab[™] (v. 7.4.0, The MathWorks[™], Natick, MA, USA).

133

134 The raw horizontal and vertical displacement time-histories from the 200 Hz cameras were
135 digitally filtered using a fourth-order Butterworth filter at 24 Hz. The filtered displacement
136 data were combined with individual-specific segmental inertia data obtained from 95 direct
137 measurements on the sprinter (Yeadon, 1990). The mass of each foot was increased by 0.2 kg
138 to account for the spiked shoes and the division of spike mass was determined directly from
139 the ratio of forefoot:rearfoot length. Whole body CM displacement time-histories were then
140 calculated using the summation of segmental moments approach (Winter, 2005). Joint angles
141 were determined and were resampled at 1000 Hz using an interpolating cubic spline before
142 their derivatives were numerically determined. Touchdown distance was determined as the
143 horizontal distance between the whole body CM and stance MTP at touchdown (recognised
144 by vertical force increasing and remaining more than two standard deviations above the zero
145 load level), with negative values indicating the MTP was behind the CM.

146

147 *Model structure*

148 A planar seven-segment angle-driven simulation model (Figure 1) was developed using
 149 Simulink[®] (v. 7.1, The Mathworks[™], Natick, MA, USA). The model incorporated a two
 150 segment representation of the stance foot (Figure 1b) due to the importance of rotation around
 151 the MTP joint in sprinting (Bezodis, Salo, & Trewartha, 2012), along with stance shank,
 152 stance thigh and swing thigh segments. The swing foot was incorporated into the swing shank
 153 segment, and the head, arms and trunk were combined into a single segment (Figure 1a). The
 154 properties of each segment were defined based on the individual-specific segmental inertia
 155 data. Segments were connected at revolute joints which permitted motion in the sagittal
 156 plane. Ground contact was modelled at each end of the forefoot segment (i.e. beneath the
 157 distal end of the toe and the MTP joint) using spring-damper systems which represented the
 158 combined visco-elasticity of the soft tissue, spiked shoe and track surface (Figure 1b). The
 159 damping terms were additionally dependent on spring length because damping increases as a
 160 spring compresses (i.e. as an increased area of the spiked shoe and track come into contact)
 161 and to avoid discontinuity in the forces at touchdown (Marhefka & Orin, 1996). Furthermore,
 162 the horizontal spring-damper systems included a term related to the vertical spring
 163 displacement because larger horizontal forces are required to achieve a given horizontal
 164 displacement when vertical spring compression is greater due to greater frictional forces
 165 (Wilson, King, & Yeadon, 2006):

$$166 \quad F_{x_i} = (-k_{x_i}x_i - c_{x_i}|x_i|\dot{x}_i)y_i \text{ (for } i = 1,2)$$

$$167 \quad F_{y_i} = -k_{y_i}y_i - c_{y_i}|y_i|\dot{y}_i \text{ (for } i = 1,2)$$

168 where F_x and F_y are the horizontal and vertical forces, x and y are the horizontal and vertical
 169 displacements relative to the original contact point, \dot{x} and \dot{y} are the derivatives of x and y , k_x
 170 and k_y , c_x and c_y are the horizontal and vertical stiffness and damping coefficients,
 171 respectively, and i represents the contact point (i.e. the toe or MTP). The model initiated at
 172 toe touchdown and forces in the MTP spring-dampers were generated when the vertical MTP

173 co-ordinate fell below a threshold level which was initially visually estimated (0.03 m) from
174 empirical data. The model terminated when the vertical toe spring returned to its initial
175 length.

176

177 ****Figure 1 near here****

178

179 *Model parameters*

180 Initial conditions for the model included horizontal and vertical velocities of the stance toe at
181 touchdown and forefoot angle and angular velocity. Each joint was angle-driven using initial
182 joint angular positions, initial angular velocities and the angular-acceleration time-histories
183 throughout stance. Similar to Wilson et al. (2006), these joint angular acceleration time-
184 histories were allowed to vary from empirical data from the instant of touchdown using a
185 combination of five sine and cosine terms (ε):

$$186 \quad \varepsilon_j(t) = j_1 \sin(t) + j_2 \cos(2t) + j_3 \sin(3t) + j_4 \cos(4t) + j_5 \sin(5t) \quad (\text{for } j = 1,6)$$

187 where j_n is the coefficient for the term of frequency n Hz at joint j . The coefficients applied
188 to the angular acceleration input parameters were allowed to vary between $\pm 1250^\circ/\text{s}^2$
189 (approximately 1% of the highest empirically recorded peak angular accelerations). Input
190 data from each empirical trial (hereafter Trial 1, 2 and 3) were separately used in matching
191 optimisations to determine the required coefficients for the toe and MTP horizontal and
192 vertical spring-damper systems and the angular acceleration functions at each joint which
193 provided the closest match between the model and each empirical trial. All horizontal foot-
194 ground interface stiffness and damping coefficients were allowed to vary between 0 and $1.0 \times$
195 10^6 N/m^2 and Ns/m^3 , respectively, and all vertical stiffness and damping coefficients between
196 0 and $1.0 \times 10^5 \text{ N/m}$ and Ns/m^2 , respectively. Similar to previous procedures (Wilson et al.,
197 2006; Yeadon & King, 2002), variation was also permitted in the touchdown estimates of

198 linear toe velocity (± 0.25 m/s), forefoot angle ($\pm 1^\circ$) and forefoot angular velocity ($\pm 25^\circ$ /s),
199 as well as the threshold level at which the MTP was deemed to have made contact with the
200 track (± 0.01 m).

201

202 ***Model evaluation***

203 A variable-step Runge-Kutte integration algorithm was used to advance the model and a
204 Latin Hypercube optimisation algorithm (McKay, Beckman, & Conover, 1979) was used to
205 find an optimum match with reality within the specified limits. The closeness of the match
206 between the model and empirical data from Trial 1 was evaluated using five kinetic and
207 kinematic criteria (Table 1) based on previous model evaluations (Yeadon & King, 2002;
208 Wilson et al., 2006; Hiley & Yeadon, 2007) as well as the specific application of this model.
209 The orientation criterion provided an additional kinematic indication of any systematic effect
210 of the cumulative configuration differences whilst the GRF accuracy criterion verified that
211 accurate impulses were not achieved as a result of large fluctuations above and below the
212 empirical GRF. The power criterion was used as an objective and appropriate measure of first
213 stance phase performance (Bezodis, Salo, & Trewartha, 2010; de Koning et al., 1992; van
214 Ingen Schenau et al., 1991; 1994). Errors in degrees were equated to those in percent (e.g.
215 Wilson et al., 2006; Yeadon & King, 2002) and the mean value of the five criteria yielded an
216 overall score reflective of the closeness of the match between the model and empirical data.
217 To quantify the robustness of the optimised foot-ground interface parameters, an independent
218 re-optimisation analysis was undertaken using the optimised spring-damper coefficients
219 obtained from the two empirical trials which were not used in the initial evaluation (i.e. Trials
220 2 and 3). These coefficients were independently determined using the same methods as
221 previously outlined, and were then used in the foot-ground interface alongside the remaining
222 input data from Trial 1, which was again allowed to vary within the previously described pre-

223 determined limits. The accuracy of this match was evaluated using the five criteria described
224 in Table 1.

225

226 ****Table 1 near here****

227

228 *Simulations using the model*

229 To determine the effects of touchdown distance on average horizontal external power
230 production during the first stance phase, the initial knee joint angle was systematically varied
231 at touchdown by $\pm 10^\circ$ in 1° increments. This varied touchdown distance from -0.9 to
232 -14.1 cm (an increasingly negative number represents the foot further behind the CM at
233 touchdown; touchdown distance was -7.3 cm in the matched optimisation). Manipulations
234 were made to the knee joint due to its aforementioned importance in the first stance phase
235 (Bezodis et al., 2014; Charalambous et al., 2012; Debaere et al., 2013), particularly in relation
236 to the lean of the body relative to the stance foot (Debaere et al., 2013). To determine the
237 effects of ankle joint dorsiflexion during early stance on average horizontal external power
238 production during the first stance phase, the ankle joint angular acceleration time-history
239 from the matched optimisation was combined with the following function:

$$\varepsilon(t) = a \cdot \cos \frac{t}{2} - b \cdot \sin \frac{t}{4}$$

240 Coefficients a and b were adjusted to yield ankle angular acceleration input data for five
241 simulations which reduced the amount of dorsiflexion during early stance without altering the
242 overall net change in ankle angle throughout stance. All simulations started from the matched
243 optimisation ankle angle of 98° and ended at the matched optimisation angle of 149° , but the
244 minimum ankle angle during stance ranged from 84° to 90° (compared to 82° in the matched
245 optimisation; see Figure 2). For all simulations, the remaining input data used the values from
246 the matched optimisation. Each simulation was advanced until the stance toe left the ground

247 and ground contact was terminated. The effect of the simulated changes on selected output
248 variables were identified to allow the primary aim of this paper to be addressed.

249

250 ****Figure 2 near here****

251

252 **Results**

253 A mean evaluation score of 5.2% was obtained between the empirical data from Trial 1 and
254 the matched optimisation. This evaluation comprised a mean difference of 5.7° in
255 configuration throughout stance (Figure 3), 8.6° in orientation throughout stance, 8.3% in the
256 overall GRF accuracy (Figure 4), 1.4% in the mean stance phase horizontal and vertical
257 impulses, and 2.0% in average horizontal external power during stance. The optimised
258 spring-damper coefficients used for modelling the foot-ground interface in each of the three
259 trials are presented in Table 2. When assessing the foot-ground interface parameters, the
260 overall evaluation scores for the two independent evaluations using the spring-damper
261 coefficients from the matching optimisations of Trials 2 and 3 with the remaining input data
262 from Trial 1 were 7.4% and 7.0%. Values for the five individual criteria from each of these
263 independent evaluations are presented in Table 3.

264

265 ****Figure 3 near here****

266 ****Figure 4 near here****

267 ****Table 2 near here****

268 ****Table 3 near here****

269

270 In the first set of simulations, a curvilinear relationship was identified between touchdown
271 distance and horizontal external power (Figure 5a). When the foot was positioned less far

272 behind the CM at touchdown than in the matched optimisation, horizontal external power
273 production progressively decreased. When the foot was positioned further behind the CM at
274 touchdown than in the matched optimisation, horizontal external power production initially
275 increased before reaching a peak value (a 0.7% increase relative to the matched optimisation
276 at 0.02 m further back). Beyond this, horizontal external power production began to decrease
277 again. The change in the ratio of force (the horizontal component of the GRF expressed as a
278 percentage of the total GRF magnitude and averaged over the entire stance phase; Morin et
279 al., 2011) associated with each of these simulations is presented in Figure 5b. In the second
280 set of simulations, as ankle joint dorsiflexion during early stance was reduced, horizontal
281 external power exponentially increased (Figure 6a). The peak resultant plantar flexor
282 moments associated with each of these simulations were extracted from the model and also
283 displayed an exponential increase (Figure 6b).

284

285 *****Figures 5a and 5b near here*****

286 *****Figures 6a and 6b near here*****

287

288

289 **Discussion and Implications**

290 This study developed and evaluated a simulation model in order to systematically determine
291 the effects of manipulations to touchdown distance and ankle joint dorsiflexion during early
292 stance on average horizontal external power production during the first stance phase. Using
293 empirical input data from an international-level sprinter, the model was quantitatively
294 demonstrated to match reality closely based on five criteria that were specific to early
295 acceleration technique and performance. The movement pattern of this sprinter during the
296 first stance phase (see sprinter B, Bezodis et al., 2014 for detailed stance leg joint mechanics

297 from the same empirical data collection) is clearly representative of that exhibited by other
298 international-level and highly-trained sprinters (e.g. Bezodis et al., 2014; Charalambous et al.,
299 2012; Debaere et al., 2013). These simulation results therefore provide preliminary evidence
300 to support and inform the design of future experimental research and applied interventions
301 (Knudson et al., 2014) with other participants from this population. However, a further
302 benefit of this novel, exploratory research is that it demonstrates that this simulation model
303 provides a means for assessing the efficacy of individual technique manipulations when
304 appropriate input parameters have been obtained. Hypothesis 1 was partly accepted as it was
305 found that positioning the stance foot increasingly further behind the CM at touchdown led to
306 small improvements in horizontal external power generation, but that this was only true up a
307 point. Beyond this, horizontal external power began to decrease again. Hypothesis 2 was
308 accepted as reductions in the range of dorsiflexion during early stance were shown to increase
309 the horizontal external power generated. This required greater peak resultant plantar flexor
310 moments which increased considerably as the reductions in dorsiflexion were systematically
311 increased.

312

313 The touchdown distance simulations revealed a curvilinear relationship with horizontal
314 external power (Figure 5a). This provided some support for the previous suggestions
315 (Bezodis et al., 2008; Kugler & Janshen, 2010) upon which hypothesis 1 was devised that
316 placing the foot further behind the CM at touchdown leads to increases in horizontal external
317 power production. However, an optimum touchdown distance was found to exist, beyond
318 which increasingly negative touchdown distances were associated with reductions in
319 horizontal external power production. A clear relationship existed between touchdown
320 distance and the ratio of force (Figure 5b): as the foot was positioned further behind the CM
321 at touchdown, the ratio of the horizontal component of the GRF to the total GRF magnitude

322 increased. The range in ratio of force values across all simulations is comparable to the range
323 exhibited during the first stance phase (approximately 35 to 55%) by the nine international-
324 and national-level sprinters analysed by Rabita et al. (2015), providing further confidence in
325 the structure and outputs of the model. It has previously been identified that a high ratio of
326 force is associated with superior sprint acceleration performance (Kugler & Janshen, 2010;
327 Morin et al., 2011; Rabita et al., 2015) and the current simulation results identify touchdown
328 distance as a specific technical factor which affects the ratio of force produced. Given that
329 horizontal external power exhibited a curvilinear relationship with touchdown distance,
330 continuing to increase the ratio of force through greater negative touchdown distances did not
331 lead to continued performance improvements. Further analysis of the model outputs revealed
332 that the magnitude of the stance-averaged resultant GRF production was less at the greater
333 negative touchdown distances where the ratio of force was highest. During sprint
334 acceleration, it has been suggested that provided sufficient vertical impulse is produced, all
335 remaining strength should be directed towards the production of propulsive horizontal
336 impulse (Hunter, Marshall, & McNair, 2005). The reduction in horizontal external power
337 production as touchdown distance became increasingly negative could therefore be reflective
338 of an inability of the sprinter to generate sufficient vertical impulse from this touchdown
339 position. It is conceivable that the body configurations at the larger negative touchdown
340 distances are associated with poor force producing capabilities *per se* but further investigation
341 is needed as factors such as specific muscle length and velocity changes cannot be accounted
342 for with the current modelling approach. Coaches and researchers should be encouraged to
343 explore strategies for manipulating foot placement with a view to finding the optimum
344 touchdown distance for a given sprinter. Although the trajectory of the CM is not visible to
345 coaches, its path during the first flight phase is fully determined at block exit. This first flight
346 phase provides sufficient duration (Bezodis, Salo, & Trewartha, 2015) for technical

347 adjustments at the leading swing knee to alter the location of the foot relative to the CM at
348 touchdown. However, caution is advised not to over-increase the negative touchdown
349 distance as placing the foot too posteriorly may be detrimental to performance.

350

351 The simulations which manipulated ankle joint dorsiflexion during early stance provided
352 support for the empirically-based assertions of Charalambous et al. (2012) and Bezodis et al.
353 (2014) upon which hypothesis 2 was devised. As the amount of dorsiflexion exhibited at the
354 ankle joint during early stance was progressively reduced, average horizontal external power
355 was found to increase exponentially (Figure 6a). Further investigation of the model outputs
356 revealed that this initially occurred due to a reduction in ground contact time without a
357 corresponding decrease in the net horizontal impulse generated. At greater reductions in
358 dorsiflexion, performance was enhanced due to both a shorter ground contact time and
359 greater net horizontal impulse generation. Reducing dorsiflexion during early stance likely
360 requires a 'stiffer' ankle joint (Charalambous et al., 2012). The peak resultant ankle plantar
361 flexor moments associated with each simulation were 7, 16, 30, 56 and 80 Nm greater than
362 the matched optimisation (299 Nm), respectively (Figure 6b). Resultant plantar flexor
363 moments have been shown to be higher in other exercises (e.g. maximal hopping, group
364 mean = 345 Nm; Farley & Morgenroth, 1999) than they are in the first stance phase.
365 Although the higher end of the current simulated increases in peak resultant plantar flexor
366 moment may therefore be an unrealistic expectation, even the smallest simulated reduction in
367 dorsiflexion was beneficial for performance and was associated with only a 7 Nm increase in
368 the peak resultant plantar flexor moment. This suggests that any reduction in ankle
369 dorsiflexion during early stance could be beneficial for horizontal external power production
370 and sprinters could therefore seek to increase their reactive plantar flexor strength through
371 plyometric training if endeavouring to improve early acceleration performance. Although

372 small reductions in ankle dorsiflexion may be difficult to accurately quantify, coaches could
373 seek to monitor this through changes to contact time or ground reaction forces given the
374 appropriate equipment.

375

376 The current overall difference between the model and empirical data (5.2%) can be
377 considered a close match compared with previously published kinetic and kinematic
378 evaluations of angle-driven models containing ground contact (e.g. 5.6 – 9.4%; Wilson et al.,
379 2006). The five individual criteria indicated that no single kinematic or kinetic aspect of the
380 model was matched considerably better than the others. The optimised foot-ground interface
381 spring-damper coefficients (Table 2) cannot be directly compared to previously published
382 angle-driven models containing ground contact due to the model-specific nature of the foot-
383 ground interface (i.e. two-segment structure of the foot, dependence of damping terms on
384 spring lengths, dependence of horizontal springs on vertical spring displacements in the
385 current model). The values obtained offer a sensible and relatively consistent representation
386 of ground contact with large horizontal forces consistently generated in the toe springs and
387 large vertical forces consistently due to the stiffness of the MTP springs once the MTP had
388 made contact with the ground (Table 2). Ultimately, the appropriateness of the foot-ground
389 interface should be considered in the context of modelled GRF profiles. The current
390 evaluation score for GRF accuracy (8.3%) compares favourably against previous angle-
391 driven models which have used a single-segment foot to model ground contact and returned
392 values of 9 to 22% using an identical GRF accuracy criterion (Gittoes et al., 2006; Wilson et
393 al., 2006). The impulse criterion scores (1.4%) further confirm the systematic closeness of the
394 current match. This good representation of the external kinetics may in part be due to the
395 novel inclusion of a two-segment foot in the current model. Such an approach could therefore

396 improve the modelling of ground contact in other activities where considerable MTP motion
397 exists but has previously been overlooked.

398

399 The variation in the optimised spring-damper coefficients between trials (Table 2) is
400 consistent with previous detailed evaluations of multiple trials. In a landing model, optimised
401 stiffness coefficients ranging from 3.9×10^5 to 1.9×10^9 N/m, and 9.5×10^4 to 2.0×10^9 N/m
402 at the toe and heel, respectively and damping coefficients ranging from 1.6×10^5 to $1.9 \times$
403 10^8 Ns/m, and 1.0×10^4 to 2.0×10^7 Ns/m at the toe and heel, respectively, were determined
404 between different trials (Gittoes, 2004). These results confirm that foot-ground interface
405 spring-damper coefficients are typically trial-specific, even when trials have been collected
406 from a single participant (Yeadon, Kong, & King, 2006). Although this highlights the need
407 for simulation models to initially be customised to an individual using empirical data from
408 specific trials, the results of the current independent re-optimisation analysis confirmed that
409 this model was relatively insensitive to these parameters (the global error of 5.2% increased
410 to a maximum of 7.4%). These independent re-optimisation results (Table 3) indicated that
411 the use of independent foot-ground interface spring-damper coefficients from different trials
412 still yielded accurate model output data. The model is clearly not overly sensitive to changes
413 in these input parameters, and the fact that none of the individual evaluation criteria increased
414 markedly more than any of the others (Table 3) as a result of these independent alterations
415 again supports the robustness of this model.

416

417 As with any theoretical investigations, by simplifying the human body into a computer-based
418 representation, several assumptions were made. The two-dimensional nature of the model is
419 consistent with the majority of empirical sprint acceleration research where sagittal plane
420 motion is of primary concern and non-sagittal forces are negligible (Debaere et al., 2013;

421 Rabita et al., 2015). The head, arms and trunk were combined in to a single segment, but
422 dividing the foot into two segments about the MTP joint helped to provide realistic
423 representations of the ground reaction forces. An angle-driven approach to actuate the model
424 was adopted due to the applied aims of this study as kinematic aspects of technique cannot be
425 directly manipulated with a torque- or muscle-driven model (Yeadon & King, 2002). This
426 approach therefore provided the most appropriate means with which to address our
427 technique-related hypotheses and is most appropriate to practical training situations in which
428 the coaching cues are generally kinematic in nature. For questions with a strength-related
429 focus, this model can now be adapted so it can be driven by joint torques and there is
430 potential to seek to develop a version driven by muscle actuators. This angle-driven model
431 provides a useful framework which can now be used to investigate the importance of other
432 aspects of technique for early acceleration performance.

433

434 **Conclusion**

435 The current study has developed, evaluated and applied a simulation model to investigate and
436 further the understanding of early acceleration technique and performance. The first set of
437 simulations extended previous empirical research which had advocated the production of a
438 more horizontally-directed GRF vector by identifying alterations to touchdown distance as a
439 means of achieving this. However, the simulation results provided preliminary evidence
440 suggesting the existence of potential limits to the benefits of positioning the foot further
441 behind the CM at touchdown and coaches should be wary of encouraging foot placement too
442 far behind the CM where performance benefits may be reduced. The second set of
443 simulations provided preliminary evidence for the beneficial effects of reducing ankle joint
444 dorsiflexion during early stance on early acceleration performance and identified the need for
445 coaches to increase ankle plantar flexor strength to facilitate this. Intervention studies are

446 required to extend these findings and to determine how coaches could affect early
447 acceleration performance through specific technical or physical training interventions related
448 to the above features.

449

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454 **References**

455 Bezodis, I. N., Kerwin, D. G., & Salo, A. I. T. (2008). Lower-limb mechanics during the
456 support phase of maximum-velocity sprint running. *Medicine & Science in Sports &*
457 *Exercise, 40*, 707-715.

458

459 Bezodis, N. E, Salo, A. I. T, & Trewartha, G. (2008). Understanding elite sprint start
460 performance through an analysis of joint kinematics. *International Society of Biomechanics*
461 *in Sports Conference Proceedings, 26*, 498-501. Retrieved from [https://ojs.uib.uni-](https://ojs.uib.uni-konstanz.de/cpa/article/view/1912/1781)
462 [konstanz.de/cpa/article/view/1912/1781](https://ojs.uib.uni-konstanz.de/cpa/article/view/1912/1781).

463

464 Bezodis, N. E, Salo, A. I. T, & Trewartha, G. (2010). Choice of sprint start performance
465 measure affects the performance-based ranking within a group of sprinters: which is the most
466 appropriate measure? *Sports Biomechanics, 9*, 258-269.

467

468 Bezodis, N. E, Salo, A. I. T, & Trewartha, G. (2012). Modeling the stance leg in 2D analyses
469 of sprinting: inclusion of the MTP joint affects joint kinetics. *Journal of Applied*
470 *Biomechanics, 28*, 222-227.

471

472 Bezodis, N. E, Salo, A. I. T, & Trewartha, G. (2014). Lower limb joint kinetics during the
473 first stance phase in athletics sprinting: Three elite athlete case studies. *Journal of Sports*
474 *Sciences, 32*, 738-746.

475

476 Bezodis, N. E., Salo, A. I. T., & Trewartha, G. (2015). Relationships between lower-limb
477 kinematics and block phase performance in a cross section of sprinters. *European Journal of*
478 *Sport Science, 15*, 118-124.

479

480 Charalambous, L., Irwin, G., Bezodis, I. N., & Kerwin, D. G. (2012). Lower limb joint
481 kinetics and ankle joint stiffness in the sprint start push-off. *Journal of Sports Sciences*, 30, 1-
482 9.

483

484 de Koning, J. J., de Groot, G., & van Ingen Schenau, G. J. (1992). A power equation for the
485 sprint in speed skating. *Journal of Biomechanics*, 25, 573-580.

486

487 Debaere, S., Delecluse, C., Aerenhouts, D., Hagman, F., & Jonkers, I. (2013). From block
488 clearance to sprint running: Characteristics underlying an effective transition. *Journal of*
489 *Sports Sciences*, 31, 137-149.

490

491 Farley, C. T., & Morgenroth, D. C. (1999). Leg stiffness primarily depends on ankle stiffness
492 during human hopping. *Journal of Biomechanics*, 32, 267-273.

493

494 Gittoes, M. J. R. (2004). *Contributions to impact loading in females during vertical drop*
495 *landings* (Unpublished doctoral dissertation). University of Bath, UK.

496

497 Gittoes, M. J. R., Brewin, M. A., & Kerwin, D. G. (2006). Soft tissue contributions to impact
498 forces simulated using a four-segment wobbling mass model of forefoot-heel landings.
499 *Human Movement Science*, 25, 775-787.

500

501 Hiley, M. J., & Yeadon, M. R. (2003a). Optimum technique for generating angular
502 momentum in accelerated backward circles prior to a dismount. *Journal of Applied*
503 *Biomechanics*, 19, 119-130.

504

505 Hiley, M. J., & Yeadon, M. R. (2003b). The margin for error when releasing the high bar for
506 dismounts. *Journal of Biomechanics*, 36, 313-319.

507

508 Hiley, M. J., & Yeadon, M. R. (2007). Optimization of backward giant circle technique on
509 the asymmetric bars. *Journal of Applied Biomechanics*, 23, 300-308.

510

511 Hunter, J. P., Marshall, R. N., & McNair, P. J. (2005). Relationships between ground reaction
512 force impulse and kinematics of sprint-running acceleration. *Journal of Applied
513 Biomechanics*, 21, 31-43.

514

515 Johnson, M. D., & Buckley, J. G. (2001). Muscle power patterns in the mid-acceleration
516 phase of sprinting. *Journal of Sports Sciences*, 19, 263-272.

517

518 Kearney, J. T. (1999). Sport performance enhancement: design and analysis of research.
519 *Medicine & Science in Sports & Exercise*, 31, 755-756.

520

521 Knudson, D., Elliott, B., & Hamill, J. (2014). Proposing application of results in sport and
522 exercise research reports. *Sports Biomechanics*, 13, 195-203.

523

524 Kugler, F., & Janshen, L. (2010). Body position determines propulsive forces in accelerated
525 running. *Journal of Biomechanics*, 43, 343-348.

526

527 Marhefka, D. W., & Orin, D. E. (1996). Simulation of contact using a nonlinear damping
528 model. *Robotics and Automation*, 2, 1662-1668.

529

530 McKay, M. D., Beckman, R. J., & Conover, W. J. (1979). A comparison of three methods for
531 selecting values of input variables in the analysis of output from a computer code.
532 *Technometrics*, 21, 239-245.

533

534 Morin, J.-B., Edouard, P., & Samozino, P. (2011) Technical ability of force application as a
535 determinant factor of sprint performance. *Medicine & Science in Sports & Exercise*, 43,
536 1680-1688.

537

538 Rabita, G., Dorel, S., Slawinski, J., Saez-de-Villarreal, E., Couturier, A., Samozino, P.,
539 Morin, J.-B. (2015). Sprint mechanics in world-class athletes: a new insight into the limits of
540 human locomotion. *Scandinavian Journal of Medicine & Science in Sports*, in press (doi:
541 10.1111.sms.12389).

542

543 van Ingen Schenau, G. J., de Koning, J. J., & de Groot, G. (1994). Optimisation of sprinting
544 performance in running, cycling and speed skating. *Sports Medicine*, 17, 259-275.

545

546 van Ingen Schenau, G. J., Jacobs, R., & de Koning, J. J. (1991). Can cycle power predict
547 sprint running performance? *European Journal of Applied Physiology and Occupational*
548 *Physiology*, 63, 255-260.

549

550 Wilson, C., King, M. A., & Yeadon, M. R. (2006). Determination of subject-specific model
551 parameters for visco-elastic elements. *Journal of Biomechanics*, 39, 1883-1890.

552

553 Winter, D. A. (2005). *Biomechanics and motor control of human movement*. (3rd ed.). New
554 York: Wiley.

555

556 Yeadon, M. R. (1990). The simulation of aerial movement 2: a mathematical inertia model of
557 the human-body. *Journal of Biomechanics*, 23, 67-74.

558

559 Yeadon, M. R., & Challis, J. H. (1994). The future of performance-related sports
560 biomechanics research. *Journal of Sports Sciences*, 12, 3-32.

561

562 Yeadon, M. R., & King, M. A. (2002). Evaluation of a torque-driven simulation model of
563 tumbling. *Journal of Applied Biomechanics*, 18, 195-206.

564

565 Yeadon, M. R., Kong, P. W., & King, M. A. (2006). Parameter determination for a computer
566 simulation model of a diver and a springboard. *Journal of Applied Biomechanics*, 22, 167-
567 176.

568 **Tables**

569 Table 1. Definition of the five criteria used to evaluate the agreement of the optimised kinetic
 570 and kinematic match between the model and empirical data.

Criterion	Definition
Configuration	The mean RMS difference between the model and empirical joint angle time-histories at each 1% of stance from the six joints.
Orientation	The RMS difference between the model and empirical HAT segment angle at each 1% of stance.
Impulse	The average percentage difference between the model and empirical vertical and net horizontal impulses from the entire stance phase
Ground reaction force accuracy	The RMS difference between the model and empirical horizontal and vertical ground reaction forces at each 1% of stance, expressed as a percentage of the total force excursion.
Power	The percentage difference between the model and empirical average horizontal external power generated during stance.

571 RMS: Root mean square; HAT: head, arms and trunk.

572 Table 2. Optimised stiffness and damping coefficients for the representation of the foot-
573 ground interface from the matching optimisations for all three trials.

Parameter	Trial 1	Trial 2	Trial 3
Horizontal toe stiffness (N/m ²)	235,850	914,274	196,843
Horizontal MTP stiffness (N/m ²)	239,990	1,327	7,031
Horizontal toe damping (Ns/m ³)	110,310	544,526	55,458
Horizontal MTP damping (Ns/m ³)	0	4,008	0
Vertical toe stiffness (N/m)	60	3,565	11,594
Vertical MTP stiffness (N/m)	48,661	35,024	28,902
Vertical toe damping (Ns/m ²)	42	46	51
Vertical MTP damping (Ns/m ²)	15,590	894	245

574

575

576 Table 3. Evaluation scores from the independent re-optimisation of Trial 1 using of spring-
577 damper coefficients from Trials 2 and 3.

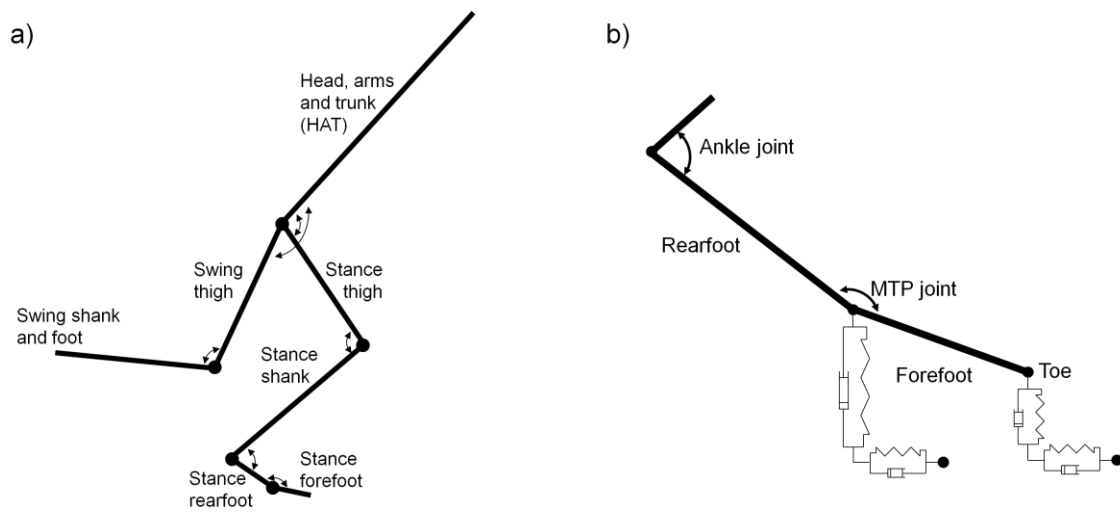
	With Trial 2 coefficients	With Trial 3 coefficients
Configuration	8.1°	7.1°
Orientation	11.5°	10.6°
Impulse	2.3%	4.6%
Ground reaction force accuracy	14.3%	9.4%
Horizontal external power	1.0%	3.3%
Mean	7.4%	7.0%

578

579

580 **Figures**

581



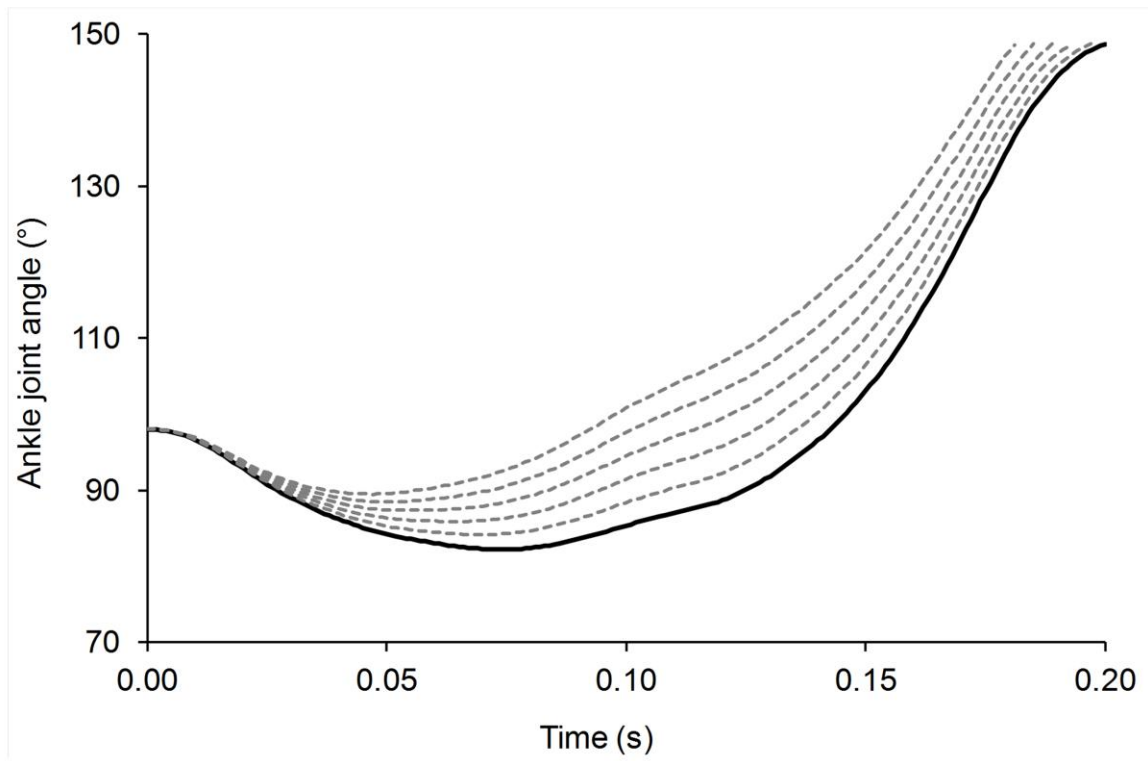
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583 Figure 1. a) Basic structure of the seven-segment simulation model. b) Structure used to

584 represent ground contact which comprised horizontal and vertical spring-damper systems

585 between the foot and the ground at the toe (distal hallux) and MTP joint.

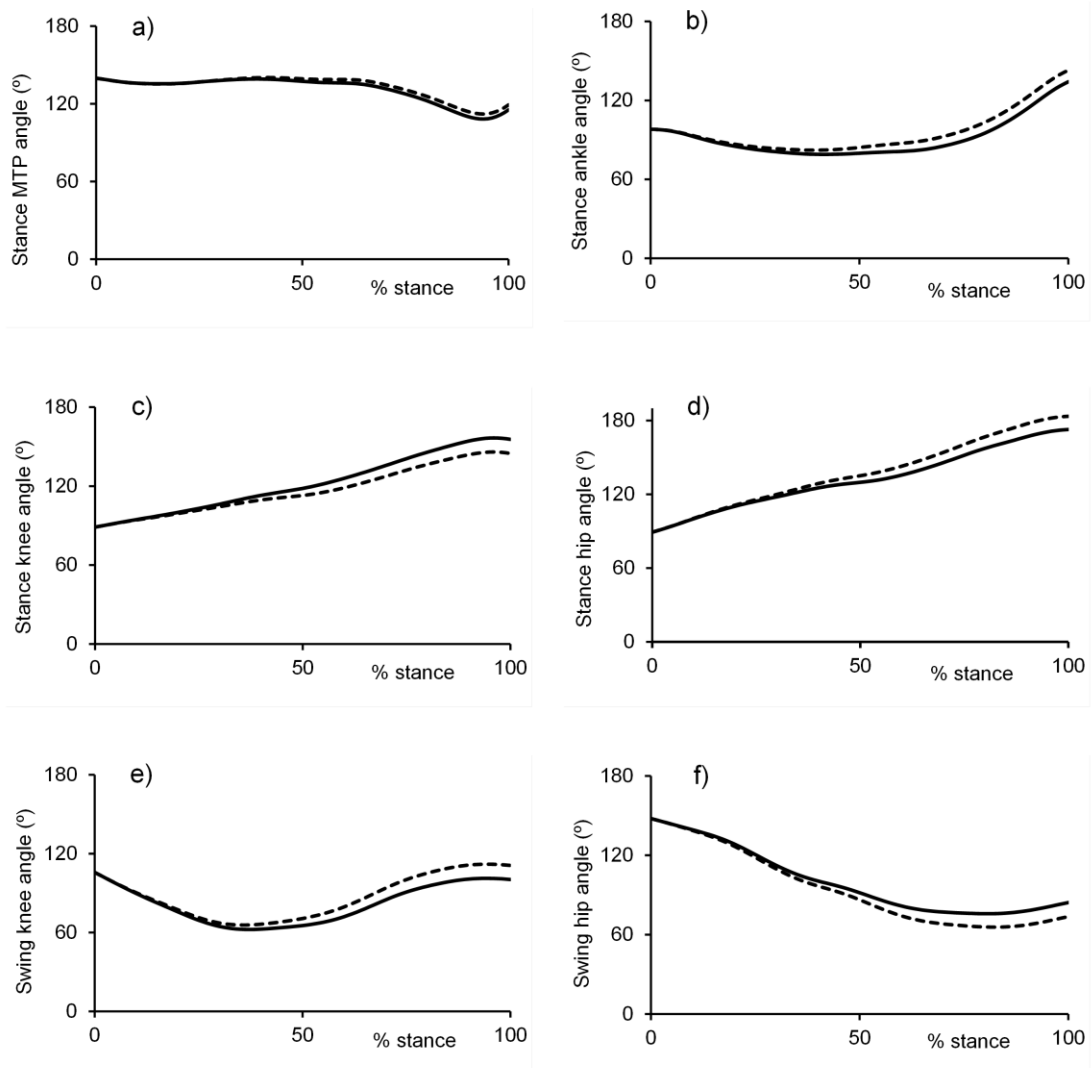
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588 Figure 2. The ankle joint angle time histories from the matched optimisation (solid black line)
589 and each of the five simulations (dotted grey lines) after addition of the respective sine and
590 cosine terms.

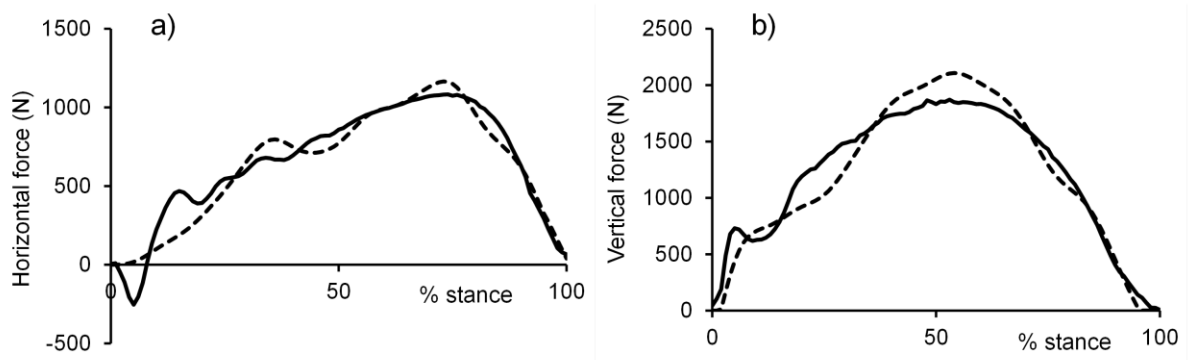
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593 Figure 3a-f. Joint angle time histories for the six angle-driven joints from the matching
 594 optimisation (empirical data = solid line, model data = dotted line).

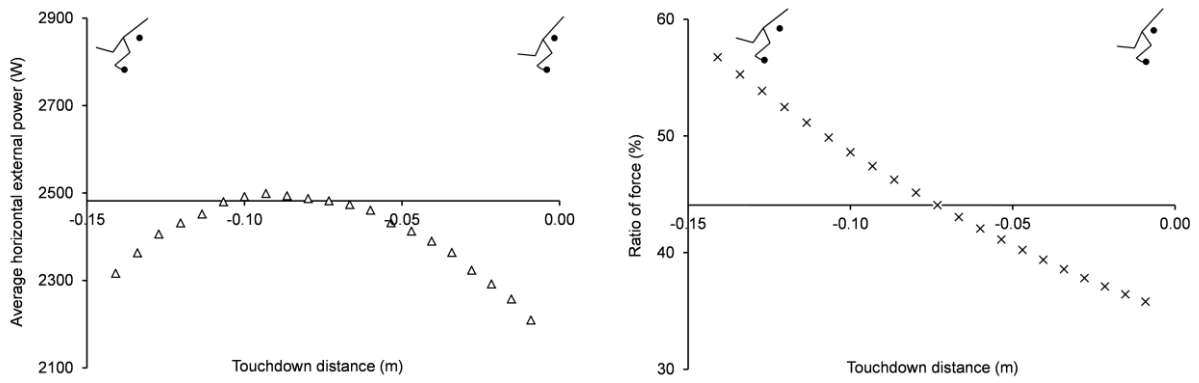
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597 Figure 4a-b. Horizontal (a) and vertical (b) ground reaction force time-histories from the
598 matching optimisation (empirical data = solid line, model data = dotted line).

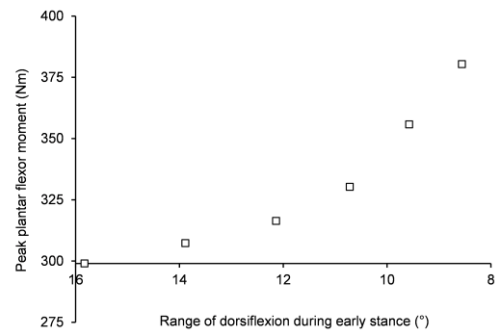
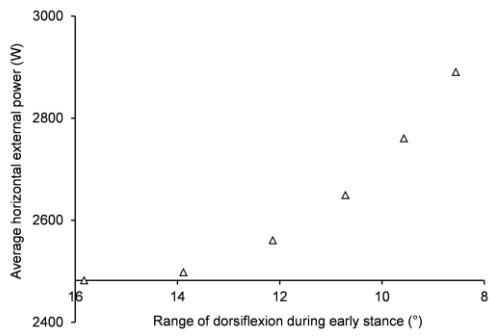
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601 Figure 5. The effect of simulated changes in touchdown distance on (a) average horizontal
 602 external power. and (b) ratio of force. The stick figures provide illustrations (not to scale) of
 603 the positions of the centre of mass and MTP joint, the horizontal distance between which is
 604 the touchdown distance (i.e. a greater negative value represents the stance toe further behind
 605 the CM at touchdown).

606



607

608 Figure 6. The effect of simulated changes in ankle joint dorsiflexion during early stance on

609 (a) average horizontal external power and (b) and peak resultant plantar flexor moment.