

Acute effects of wearable thigh and shank loading on spatiotemporal and kinematic variables during maximum velocity sprinting

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1 **Abstract**

2 Light wearable resistance is used in sprint training, but the scientific evidence to guide its
3 implementation is limited. This study investigated thigh and shank loading protocols which
4 were matched based on the average increase in moment of inertia about the hip over a stride
5 cycle. Seven university-level sprinters completed three counterbalanced conditions (unloaded,
6 shank-loaded, thigh-loaded), and kinematic variables were measured between 30 and 40 m.
7 Both thigh and shank loading led to small reductions in step velocity (mean change = -1.4%
8 and -1.2%, respectively). This was due to small reductions in step frequency (-1.8%; -1.7%)
9 because of small increases in contact time (+2.7%; +1.5%) in both conditions as well as a small
10 increase in flight time (+2.0%) in the shank-loaded condition. Both conditions led to moderate
11 increases in hip extension at toe-off (+2.7°; +1.4°), whilst thigh loading led to a small reduction
12 in peak hip flexion angle during swing (-2.5°) and shank loading led to a small increase in peak
13 biceps femoris muscle-tendon-unit length (+0.4%). Thigh and shank loading can both be used
14 to provide small reductions in sprint velocity, and each have specific overload effects which
15 must be considered in the rationale for their implementation.

16 **Introduction**

17 Sprint training is typically periodised by including highly-specific resistance training when
18 athletes are undertaking specialised developmental exercises as they transition towards
19 competition (Bompa, 1999; Bondarchuk, 2006; Wild, Bezodis, Blagrove, & Bezodis, 2011).
20 Because highly-specific resistance training is important for the effective transfer of strength to
21 sprinting performance (Delecluse, 1997; Young, 2006; Cronin, Ogden, Lawton, & Brughelli,
22 2007), sprint training with added resistance is common practice. For example, bands,
23 parachutes, sleds, and weighted belts and vests are often advocated as ways of providing
24 specific overload during sprints, and the specificity and efficacy of such methods has been the
25 focus of considerable research (e.g. Spinks, Murphy, Spinks, & Lockie, 2007; Alcaraz, Palao,
26 Elvira, & Linthorne, 2008; Cronin, Hansen, Kawamori, & McNair, 2008; Clark, Stearne, Walts,
27 & Miller, 2010). Whilst these methods provide an additional external force directly to the torso,
28 the direct application of light wearable masses to lower-body segments has also been
29 investigated (Ropret, Kukolj, Ugarkovic, Matavulj, & Jaric, 1998; Bennett, Sayers, & Burkett,
30 2009; Pajić, Kostovski, Ilić, Jakovljević, & Preljević, 2011; Simperingham & Cronin, 2014;
31 Macadam, Simperingham, & Cronin, 2017a; Macadam et al., 2019). This is proposed to more
32 specifically challenge the rotational capabilities of the legs due to an increase in their moment
33 of inertia, potentially making the overload more specific to sprinting (Macadam, Cronin, &
34 Simperingham, 2017b).

35

36 Whilst lower-body light wearable resistance is not a new concept, the scientific evidence behind
37 its effects on maximum velocity sprinting remains limited. This is partly because of the
38 flexibility in how lower-body light wearable resistance can be applied which has led to small,
39 but important, differences in the location and magnitude of load application between studies.
40 For example, when a 10% segmental mass increase was applied to both the shank and thigh
41 (approximately 3% body mass (BM) in total), a significant reduction in average stride velocity

42 (-4.7%) was reported between 25 and 30 m (Bennett et al., 2009). This was accompanied by a
43 non-significant reduction in stride frequency (-2.2%) and a non-significant increase in contact
44 time (+8.9%), with stride length effects not reported. Bennett et al. (2009) also observed a
45 reduction in peak hip flexion during swing when loaded, but no acute effects on hip angle during
46 stance or on knee angle during any measured points in the stride cycle. Simperingham and
47 Cronin (2014) also simultaneously loaded the thigh and shank (5% increase in BM in total) and,
48 using a non-motorised treadmill, observed a significant reduction in peak velocity (-4.9%). As
49 there was no significant change in step length, the reduction was due to decreases in step
50 frequency (-3.5%) which were associated with significant increases in contact time (+4.3%).

51

52 Where loads have been added to either the thigh *or* shank segments, gradual increases in shank
53 loading (from 0.6 to 1.2 to 1.8 kg distally on each shank, i.e. up to 4.8% BM) have been shown
54 to progressively decrease average velocity by up to 12.8% during the 15-30 m section of a
55 maximal effort sprint (Ropret et al., 1998). Similar to the general effects observed with
56 combined shank and thigh loading (Bennett et al., 2009; Simperingham & Cronin, 2014), these
57 reductions in velocity were accompanied by reductions in stride frequency, but no significant
58 change in stride length. The addition of lighter masses to the shank (15% of segment mass, i.e.
59 ~0.37 kg per shank) has also been shown to significantly reduce velocity (-2.2%) through
60 significant increases in contact time and no change in step length (Zhang et al., 2019). When
61 load has been applied distally on the thigh (2% BM; Macadam et al., 2019), moderate reductions
62 in velocity (-2.0%) were observed during steps 15 to 23 of a maximal effort sprint. These were
63 again associated with a small reduction in step frequency (-1.8%) and no clear change in step
64 length (-0.5%), with the step frequency effects primarily due to a small increase in contact time
65 (+2.9%).

66

67 Given the variety of load locations and magnitudes used between studies, the evidence to guide
68 the applied implementation of light lower-body wearable resistance remains limited and the
69 specific prescription by practitioners therefore remains largely intuition-based. Relatively
70 heavy masses have also been used which restricts the transfer of evidence to applied practice in
71 track and field. This is because there is resistance to the use of such masses when aiming to
72 facilitate transfer during specialised developmental exercises due to the unknown effects on
73 sprinting kinematics, which could affect not only performance but also the potential risk of
74 injury. As there is a high hamstring strain injury incidence in sprinters (D'Souza, 1994; Yeung,
75 Suen, & Yeung, 2009) which may be related to muscle-tendon unit (MTU) strain during the
76 late swing phase (Kenneally-Dabrowski et al., 2019), it is also important to consider any
77 potential effects on hamstring strain.

78

79 Whilst there are clearly numerous possible combinations of load magnitudes and placements
80 both within and between segments, an important first challenge is to better understand how the
81 application of light wearable resistance to *either* the thigh segment *or* the shank segment affects
82 performance and key technical variables when compared with unloaded sprinting through a
83 direct comparison. The aim of this study was therefore to quantify the acute effects of the
84 addition of light wearable loads to either the thigh or shank during maximum velocity sprinting
85 on spatiotemporal characteristics, hip and knee joint angles at key events, and peak hamstring
86 MTU lengths. It was hypothesised that, when compared with unloaded sprinting, 1) both thigh
87 and shank loading will lead to reductions in step velocity due to reductions in step frequency
88 associated with an increase in contact time, and 2) that the effects on hip, knee and hamstring
89 MTU kinematics will differ between thigh and shank loading.

90

91 **Materials and methods**

92 *Participants*

93 Seven university-level sprinters (six male, one female; mean \pm SD: age = 21 ± 1 years; height
94 = 1.73 ± 0.09 m; mass = 71.1 ± 6.6 kg, season's best sprint time, male = 11.61 ± 0.39 s, female
95 = 12.0 s) provided written informed consent to participate in this study. All procedures were
96 approved by the Swansea University College of Engineering Research Ethics and Governance
97 Committee.

98

99 ***Data collection***

100 Data collection took place at an indoor track and participants wore tight-fitting shorts, a vest
101 top and their own spiked shoes. After completing their typical warm-up for a maximum velocity
102 session, all participants performed six 40 m sprints from a two-point start. This comprised two
103 unloaded sprints, two thigh-loaded sprints (+0.6 kg per leg) and two shank-loaded sprints (+0.2
104 kg per leg) in a counterbalanced order between participants. Participants had at least two
105 minutes of recovery between sprints within each condition, and at least five minutes between
106 conditions. The specific masses applied (Lila™ Exogen™, Sportboleh Sdh Bhd, Malaysia)
107 were selected in an attempt to provide a similar increase in moment of inertia about the hip joint
108 during an entire stride cycle (see *Determination of participant-specific wearable resistance*
109 *loads*), and to ensure that relatively light total loads (< 1% BM) were applied to the shanks in
110 order to increase the relevance to applied practice for use during specialised developmental
111 training.

112

113 An optical measurement system with infra-red light barriers (Optojump, Microgate, Italy) was
114 placed either side of the sprint lane between 30 and 40 m to obtain spatiotemporal
115 characteristics. A digital video camera (PXW-Z150, Sony, Japan) was set up perpendicular to
116 the sprint lane at the 35 m mark (sampling frequency = 120 Hz, shutter speed = $1/725$ s,
117 resolution = 1920×1080 pixels). The camera was positioned 16 m from the centre of the lane,
118 viewing the left side of the participants within a field of view approximately 10 m wide to

119 ensure that one complete stride cycle of the left leg was captured for all trials. An 8×2 m plane
120 was calibrated in the centre of the lane between the 31 and 39 m marks.

121

122 *Determination of participant-specific wearable resistance loads*

123 The knee joint angles of eight national level sprinters during a maximum velocity stride cycle
124 were manually digitised (Quintic v.29, U.K.) from the figure presented by Zhong, Fu, Wei, Li,
125 & Liu (2017). These were imported to Matlab (R2017b, MathWorks, USA) and padded via 10-
126 point reflection at both ends (Smith, 1989) low-pass filtered at 10 Hz, and resampled at every
127 1% of the stride cycle using an interpolating cubic spline. The mean knee angle during a
128 maximum velocity stride cycle was then determined (108.3°).

129

130 To ensure relevance to applied practice, it was decided that all participants would have 0.2 kg
131 added to each shank during the shank-loaded condition. Using the parallel axis theorem, the
132 mean knee angle (108.3°) determined from Zhong et al. (2017), and the segmental inertia
133 parameters of de Leva (1996), a mass of 0.6 kg on each thigh was initially determined as
134 providing comparable moment of inertia demands about the hip joint (approximately +4.5%)
135 between both loaded conditions when averaged across the stride cycle, and that this would be
136 achievable for all participants in the current study given their height and body mass. The directly
137 measured thigh and shank lengths of each participant were then used along with each
138 participant's mass and the segmental inertia parameters of de Leva (1996) to determine the
139 participant-specific percentage distances (to the nearest whole percent) along each of the thigh
140 and shank segments at which to place the centre of mass of the added loads in order to yield
141 these increases in moment of inertia. Thigh loads were placed anteriorly at $76 \pm 4\%$ (range =
142 69 to 80%) of the distance from the proximal end of the segment across the studied group, and
143 shank loads were placed anteriorly at $62 \pm 11\%$ (range = 42 to 78%) of the distance from the
144 proximal end. This yielded increases of $4.48 \pm 0.03\%$ in the moment of inertia of the leg about

145 the hip joint in the thigh-loaded condition, and $4.49 \pm 0.01\%$ in the shank-loaded condition (at
146 knee angles of 108.3°). The rotational demands about the hip joint were therefore considered
147 matched across an entire stride cycle between the two experimental conditions, with both $\sim 4.5\%$
148 higher than in the unloaded condition.

149

150 *Data analysis*

151 For all trials, a stride cycle (from left foot contact to next left foot contact) which occurred
152 between 30 and 40 m was identified (if two complete left leg strides were completed within the
153 capture volume, the one closest to the centre was used). From the optical measurement system,
154 raw values for step length, contact time and flight time were extracted. Step frequency was
155 calculated as the inverse of the sum of contact time and flight time, and step velocity was
156 calculated as the product of step length and step frequency. The step characteristics for the two
157 steps within the analysed stride were averaged to yield a single value for each variable for each
158 trial.

159

160 The raw video files were calibrated, after which a 6-point model (neck (mid C7-larynx), left
161 hip, knee and ankle joint centres, left calcaneus and left 5th metatarsal) was manually digitised.
162 Each trial was digitised twice from 10 frames prior to the initial touchdown until 10 frames
163 after the next touchdown. All raw co-ordinate time-histories were then exported for subsequent
164 analysis in Excel (Microsoft, USA) and Matlab. The mean co-ordinates across both digitisations
165 were calculated and used for all subsequent analyses to reduce the potential random error
166 associated with manual digitisation. Hip, knee and ankle joint angles were calculated at each
167 frame, and MTU lengths of the biceps femoris long head (BF_{lh}; chosen due to hamstring strain
168 injuries primarily affecting this muscle; Koulouris & Connell, 2003; Askling, Tengvar, Saartok,
169 & Thorstensson, 2007) were determined from the hip and knee flexion angles using the
170 regression equations of Hawkins and Hull (1990). The raw joint angles and MTU lengths were

171 low-pass filtered using a 4th order Butterworth filter at 10 Hz, after which the padding frames
172 were removed. All variables were then sampled at 101 evenly-spaced data points using an
173 interpolating cubic spline to represent each 1% of the stride cycle from the first left foot
174 touchdown (0%) to the next left foot touchdown (100%). The percentage of the stride cycle at
175 which specific events (maximum knee flexion during ground contact, toe-off, maximum knee
176 flexion during swing, maximum hip flexion during swing) occurred were then identified.
177 Discrete values of the joint angle and MTU data at each event were extracted, and data were
178 averaged across the two trials in each condition for each participant to obtain the dependent
179 variables used for statistical analysis.

180

181 *Statistical analysis*

182 Given the intended high specificity of the loaded conditions to unloaded sprinting and the fact
183 that previous studies with 2% BM increases on the thigh have observed moderate but non-
184 significant effects on spatiotemporal variables during sprinting (Macadam et al., 2019), both
185 traditional statistics and a magnitude-based decision approach (Batterham & Hopkins, 2006;
186 Hopkins, 2019) were used to compare each of the loaded conditions with unloaded sprinting.
187 Firstly, a repeated measures ANOVA (SPSS v. 26, IBM, USA) was conducted to determine if
188 there was a significant ($p < 0.05$) main effect of condition. In such instances, pairwise
189 comparisons were then conducted between each of the loaded conditions and the unloaded
190 condition using Fisher's LSD. Secondly, effect sizes (d ; Cohen, 1988) and their 95%
191 compatibility intervals (Hopkins, 2019) were calculated between each of the loaded conditions
192 and the unloaded condition. Thresholds of 0.2, 0.5 and 0.8 were used to define small, moderate
193 and large mean effects, respectively (Cohen, 1988). Based on a smallest worthwhile effect size
194 of 0.2 (Hopkins, 2004; Winter, Abt, & Nevill, 2014), clear effects were identified where the
195 95% compatibility interval did not overlap an effect size of both +0.2 and -0.2. The percentage
196 likelihoods of a negative | trivial | positive effect were also calculated (Batterham & Hopkins,

197 2006). For clarity, any significant main effects are explicitly reported in the written text and
198 any clear effects are described in the written results by reporting the magnitude threshold (e.g.
199 small, moderate, large). The likelihoods of the clear effects (including the qualitative
200 descriptors) are presented in the tables.

201

202 **Results**

203 There was a significant main effect of condition on step velocity, with a small reduction in step
204 velocity in both loaded conditions compared with the unloaded condition (Table 1). These
205 reductions were associated with small reductions in step frequency in both conditions, and
206 trivial or unclear increases in step length, compared with the unloaded condition (Table 1).
207 There was a significant main effect of condition on contact time, with a small increase in contact
208 time in both loaded conditions when compared with the unloaded condition. There was also a
209 small increase in flight time in the shank-loaded condition compared with the unloaded
210 condition, but the effect of the thigh-loaded condition on flight time was unclear (Table 1).

211

212 ****Table 1 near here****

213

214 There was a small reduction in hip flexion angle at touchdown and at the instant of maximum
215 hip flexion during swing in the thigh-loaded condition compared with the unloaded condition,
216 but there were no clear differences in hip angle between the shank-loaded condition and
217 unloaded condition at these instants (Table 2). At toe-off, there was a large increase in hip
218 extension angle in the thigh-loaded condition, and a moderate increase in the shank-loaded
219 condition, when compared with the unloaded condition (Table 2). At the knee joint, there were
220 unclear or trivial effects of both loaded conditions at all instances except for at maximum knee
221 flexion during swing in the thigh-loaded condition where there was a possible reduction in knee
222 flexion, although the mean effect size was less than small (Table 3).

223

224 ****Table 2 near here****

225 ****Table 3 near here****

226

227 The peak BFlh MTU length exhibited a small increase (of approximately an additional 0.4% of
228 its resting length) in the shank-loaded condition compared with the unloaded condition but there
229 was no clear effect between the thigh-loaded and unloaded conditions (Table 4). There was no
230 clear difference in the time occurrence of the peak BFlh MTU length in either experimental
231 loaded condition compared with the unloaded condition (Table 4).

232

233 ****Table 4 near here****

234

235 **Discussion and implications**

236 We aimed to quantify the acute effects of adding light wearable loads to either the thigh or
237 shank during maximum velocity sprinting on spatiotemporal characteristics, hip and knee joint
238 angles at key events, and peak hamstring MTU lengths. There were small reductions in step
239 velocity in both the thigh-loaded (mean = -1.4%) and shank-loaded (-1.2%) conditions
240 compared with the unloaded condition (Table 1). This aligned logically with previous research
241 as Macadam et al. (2019) and Zhang et al. (2019) observed slightly greater (2.0 and 2.2%,
242 respectively) reductions in velocity with respective thigh (~0.67 kg per leg) or shank (~0.37 kg
243 per leg) loads which were slightly heavier than those used in the current study. The mean
244 reduction in velocity in our shank-loaded condition was only 0.02 m/s less than that observed
245 in our thigh-loaded condition, despite the added mass being three times less. This adds support
246 to the rotational nature of light wearable resistance lower-limb overload rather than it simply
247 being an increased total system mass for the athlete to overcome.

248

249 In both conditions, the reductions in velocity were associated with small reductions in step
250 frequency, and unclear or trivial effects on step length (Table 1). This is consistent with
251 previous studies which have used a variety of loading locations and magnitudes and have
252 studied effects at a range of sprint distances (Ropret et al., 1998; Macadam et al., 2017a; 2019;
253 Simperingham & Cronin, 2014; Zhang et al., 2019), and our first hypothesis was thus accepted.
254 Step frequency reduced due to small increases in contact time in both conditions, and in flight
255 time in the shank-loaded condition (Table 1), broadly consistent with Macadam et al. (2019)
256 and Zhang et al. (2019). Our assessment of both conditions on the same participants with
257 matched rotational demands provides new evidence to suggest that shank loading may affect
258 the temporal mechanics slightly differently from thigh loading.

259

260 The increases in contact time likely occurred because of the need for a greater vertical impulse
261 to overcome the greater system mass, as evidenced by Macadam et al. (2019). As the
262 participants were presumably already producing their maximum force output in the time
263 available when unloaded, the only way to increase impulse was therefore through increased
264 contact time. This also explains why the effect was greater and clearer in the thigh-loaded than
265 the shank-loaded condition, because the increase in total system mass (and therefore the
266 required increase in impulse) in the thigh-loaded condition (+1.2 kg) was three times greater
267 than in the shank-loaded condition (+0.4 kg). The reasons for the increase in flight time in the
268 shank-loaded condition are less clear. It is possible that the contact time increase (+1.5%,
269 compared with +2.7% in the thigh-loaded condition) led to a greater than necessary increase in
270 vertical impulse given that the added mass was three times less in the shank-loaded than in the
271 thigh-loaded condition, and this thus led to an increased time subsequently spent in flight.
272 However, this did not lead to an increase in step length and thus future research is warranted to
273 further explore this.

274

275 In both loading conditions, the hip was more extended at toe-off than in the unloaded condition
276 (Table 2). Whilst few of comparable studies have reported joint kinematics, Zhang et al. (2019)
277 also observed a mean increase of 1.3° in hip extension at toe-off compared with unloaded
278 sprinting, although this was not significantly different. Greater hip extension at toe-off with
279 wearable lower-limb resistance is likely related to the aforementioned desire to maintain
280 vertical impulse and, by increasing contact time, the participants are also increasing the time
281 available for joint rotation. Whilst it has been suggested these longer contact times may relate
282 to touchdown mechanics and braking effects (Macadam et al., 2019), our results suggest that
283 the longer contact times are associated with a greater hip extension towards the end of the
284 stance-phase. Our second hypothesis was accepted as there were differences in the responses
285 between the thigh and shank loading conditions, with the thigh-loaded condition also affecting
286 the hip angle at touchdown and at maximum flexion during swing in addition to the above
287 effects at toe-off. When thigh-loaded, the hip was in a more extended/less flexed position at all
288 three events (Table 2). The reduction in maximum hip flexion is consistent with the findings of
289 Bennett et al. (2009) who presented consistent decreases in peak hip flexion when loaded for
290 all eight of their participants. Bennett et al. (2009) loaded both the thigh and shank
291 simultaneously and, when considered in the context of our findings, it is likely that the thigh
292 loading primarily led to their observed effect. Our results therefore suggest that the addition of
293 light wearable resistance to the thigh provides a specific resistance to achieving ‘front-side
294 mechanics’ (leg actions occurring in front of the extended line of the torso) about the hip. This
295 may be of interest to practitioners who place importance on ‘front-side mechanics’ and may
296 wish to provide a specific overload, although it must be acknowledged that kinematic variables
297 associated with ‘front-side mechanics’ were not related to maximum velocity in a group of
298 sprinters (mean 100 m personal best =10.86 s; Haugen et al., 2018).

299

300 Shank loading did not affect hip angles at maximum hip flexion or touchdown, and this may be
301 because the knee is more flexed during mid-swing than it is when averaged across the stride
302 cycle (as used to inform the current protocol; Zhong et al., 2017). The moment of inertia of the
303 leg about the hip joint would therefore have been relatively lower during this part of the stride
304 cycle. Shank loading can therefore be used to have less of an overload effect during mid-swing
305 and a greater effect at times when the knee is more extended, such as stance, and practitioners
306 should be cognisant of this important consideration when prescribing the location and amount
307 of load applied. For a given increase in the average rotational demands about the hip joint over
308 a whole stride cycle (as currently investigated), shank loading would yield greater overload
309 when the ground reaction force creating requirements are high during stance, whereas thigh
310 loading would yield greater overload on the hip flexors during swing. A combination of both
311 loading schemes could be used as part of a programme to provide comparable overall lower
312 body rotational overloads, but with variation in the specific mechanics affected and the required
313 intermuscular coordination patterns, so that the same structures and movements are not
314 continually overloaded in all sessions.

315

316 There were no clear effects of either loading condition on knee angle at any of the three events
317 studied during the stance phase (Table 3). Although these effects were unclear, the direction of
318 the mean differences in the shank-loaded condition opposed those which occurred in the thigh-
319 loaded condition (i.e. more flexed knee at touchdown and more extended knee at toe-off). Based
320 on the relatively greater moment of inertia effects in the shank-loaded condition during late-
321 swing and stance as discussed above, shank loading could lead to different effects during the
322 stance phase. Zhang et al. (2019) observed a reduction in knee flexion at touchdown with ~0.37
323 kg loading per shank, but the lighter shank loads purposefully used in the current study may not
324 have been sufficient to yield clear effects. During the swing phase, there was a possible
325 reduction in maximum knee flexion in the thigh-loaded condition, but the mean effect size was

326 less than small ($d = 0.19$). This may again be due to the relatively greater moment of inertia
327 about the hip in the thigh-loaded than shank-loaded condition during the swing phase, but the
328 reason why this might have affected knee flexion is not clear. One possible explanation is that
329 this is a function of the thigh loading inhibiting hip flexion during swing (as discussed above),
330 and thus the thigh segment reaches a less horizontal orientation which has a consequent effect
331 on the knee angle between this and the shank.

332

333 There was a small increase in the peak MTU length of the BFlh in the shank-loaded condition
334 compared with the unloaded condition, but there was no clear effect in the thigh-loaded
335 condition (Table 4). Given that the MTU length is a function of hip and knee kinematics, this
336 is likely explained by the lesser peak hip flexion in the thigh-loaded condition, and thus during
337 late swing when the MTU reached its peak length, the hip was in a slightly less flexed position
338 in the thigh-loaded condition. Shank loading therefore appears to lead to a small overload in
339 peak BFlh MTU length, whereas the effects of thigh loading are unclear. Finally, whilst there
340 was no clear effect of either condition on the timing of this peak MTU length, the mean size of
341 the effect ($d = 0.74$) in the shank-loaded condition was moderate and may warrant further direct
342 exploration in future research.

343

344 Whilst one limitation of our study is that we did not compare the direct effect of different
345 placements of the same absolute load, our study developed and described a novel objective
346 method for matching the rotational demands about the hip joint between different loading
347 configurations. This enabled us to assess the effects of two loading schemes which were
348 theoretically matched for the overall rotational demands across an entire stride cycle, rather
349 than observing likely increased effects with shank loading if matching the masses applied.
350 Researchers should carefully consider the design of the loading protocols (e.g. matched total
351 mass when greater shank overload is intended versus lower shank masses when matched

352 rotational demands are intended) depending on their specific question. Further limitations relate
353 to the use of two-dimensional motion analysis, as well as an optical measurement system for
354 determining step characteristics which likely led to a small over-estimation in contact time and
355 an under-estimation in flight time compared with previous research which has used force
356 platforms (e.g. Macadam et al., 2019), but these effects were consistent across all studied
357 conditions and thus do not limit our comparisons. Finally, our results are also from a relatively
358 small sample of university-level sprinters and further investigations are required to assess the
359 generalisability of these findings. Future research should also consider the acute neuromuscular,
360 physiological and endocrine responses to training with light wearable resistance so that such
361 sessions can be best programmed. This will also help to inform the planning of longer-term
362 training interventions which are ultimately required to assess whether training with light
363 wearable resistance can enhance sprinting performance.

364

365 **Conclusion**

366 Light wearable resistance applied to either the shank or thigh provides a small overload effect
367 on maximum velocity which occurs through reductions in step frequency. This is due to small
368 increases in contact time when thigh-loaded, and to small increases in both contact and flight
369 time when shank-loaded. Important to note is that one-third as much mass was applied to each
370 shank compared with each thigh segment, and thus lighter loads can be used more distally to
371 create similar performance overload effects due to the increased rotational demands associated
372 with the location of these loads. Whilst both thigh and shank loading led to increases in hip
373 extension at toe-off, thigh loading affected hip joint mechanics at other events in the stride
374 cycle, most notably in limiting the maximum hip flexion achieved during the swing phase.
375 Shank loading may provide greater relative overload effects during stance and led to small
376 increases in peak BFlh MTU length, and thus different loading locations can be used if specific
377 kinematic responses are desired.

378

379

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384

385

386 **Declaration of interest statement**

387 John Cronin is the Head of Research for Lila™ but had no role in the design of this study. All
388 other authors declare no conflict of interest.

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480 **Tables**

481

482 Table 1. Comparison of step characteristics for all three conditions.

	Main effect (p)	Thigh-loaded versus unloaded				Shank-loaded versus unloaded		
		Unloaded Mean ± SD	Thigh Mean ± SD	Shank Mean ± SD	ES ± 95% CI	Percentage likelihood of negative trivial positive effect	ES ± 95% CI	Percentage likelihood of negative trivial positive effect
Step velocity (m/s)	0.027	9.14 ± 0.44	9.01 ± 0.43*	9.03 ± 0.46	-0.26 ± 0.11^^	88 12 0	-0.22 ± 0.28^	56 43 1
Step length (m)	0.815	2.05 ± 0.13	2.06 ± 0.12	2.06 ± 0.14	0.04 ± 0.26	3 88 9	0.07 ± 0.18 ^{††}	1 93 7
Step frequency (Hz)	0.131	4.47 ± 0.31	4.39 ± 0.24	4.39 ± 0.32	-0.24 ± 0.28^	63 36 0	-0.23 ± 0.24^	60 40 0
Flight time (s)	0.501	0.111 ± 0.009	0.112 ± 0.007	0.113 ± 0.010	0.06 ± 0.47	11 64 25	0.21 ± 0.37^	2 45 53
Contact time (s)	0.022	0.114 ± 0.007	0.117 ± 0.008*	0.115 ± 0.008	0.35 ± 0.28^^	0 11 88	0.20 ± 0.29^	1 50 49

483 SD = standard deviation; ES = effect size (Cohen's d); CI = compatibility interval.

484 Where there was a significant main effect of condition (p < 0.05), significant (p < 0.05) pairwise differences for each experimental condition versus the unloaded
485 condition are notated next to the condition Mean ± SD with an asterisk.

486 ^ clear difference versus the unloaded condition (^ = possible, ^^ = likely, ^^ = very likely).

487 † trivial difference versus the unloaded condition († = possibly, †† = likely, ††† = very likely).

488 Percentage likelihoods are presented to the nearest whole number and thus the negative | trivial | positive effects may not always add up to 100%.

489 Table 2. Comparison of hip joint angles at selected discrete events for all three conditions.

Event	Main effect (p)	Unloaded Mean ± SD	Thigh Mean ± SD	Shank Mean ± SD	Thigh-loaded versus unloaded		Shank-loaded versus unloaded	
					ES ± 95% CI	Percentage likelihood of negative trivial positive effect	ES ± 95% CI	Percentage likelihood of negative trivial positive effect
Touchdown (°)	0.751	40.3 ± 3.9	39.3 ± 4.4	39.9 ± 4.9	-0.21 ± 0.35 [^]	53 46 1	-0.08 ± 0.85	37 40 22
Toe-off (°)	0.067	-13.3 ± 2.2	-16.0 ± 3.3	-14.7 ± 1.5	-0.95 ± 0.95 ^{^^}	95 4 1	-0.50 ± 0.52 ^{^^}	90 9 1
Maximum hip flexion during swing phase (°)	0.117	71.4 ± 4.8	68.9 ± 4.3	69.4 ± 4.8	-0.48 ± 0.30 ^{^^^}	97 3 0	-0.38 ± 0.61	75 22 3

490 SD = standard deviation; ES = effect size (Cohen's d); CI = compatibility interval.

491 Where there was a significant main effect of condition ($p < 0.05$), significant ($p < 0.05$) pairwise differences for each experimental condition versus the unloaded
492 condition are notated next to the condition Mean ± SD with an asterisk.

493 [^] clear difference versus the unloaded condition ([^] = possible, ^{^^} = likely, ^{^^^} = very likely).

494 [†] trivial difference versus the unloaded condition ([†] = possibly, ^{††} = likely, ^{†††} = very likely).

495 Percentage likelihoods are presented to the nearest whole number and thus the negative | trivial | positive effects may not always add up to 100%.

496 Table 3. Comparison of knee joint angles at selected discrete events for all three conditions.

Event	Main effect (p)				Thigh-loaded versus unloaded		Shank-loaded versus unloaded	
		Unloaded Mean ± SD	Thigh Mean ± SD	Shank Mean ± SD	ES ± 95% CI	Percentage likelihood of negative trivial positive effect	ES ± 95% CI	Percentage likelihood of negative trivial positive effect
Touchdown (°)	0.774	26.4 ± 6.3	25.4 ± 6.3	26.9 ± 6.1	-0.13 ± 0.59	40 50 11	0.07 ± 0.82	23 42 35
Maximum knee flexion during stance (°)	0.569	40.6 ± 6.6	38.9 ± 7.0	40.2 ± 3.8	-0.26 ± 0.57	59 36 5	-0.06 ± 0.63	30 52 18
Toe-off (°)	0.561	22.2 ± 7.1	22.8 ± 7.4	21.3 ± 6.4	0.07 ± 0.17 ^{††}	0 94 5	-0.11 ± 0.48	33 59 8
Maximum knee flexion during swing (°)	0.121	137.9 ± 7.9	136.1 ± 7.6	137.9 ± 8.9	-0.19 ± 0.21 [^]	47 53 0	0.01 ± 0.19 ^{†††}	2 96 2

497 SD = standard deviation; ES = effect size (Cohen's d); CI = compatibility interval.

498 Where there was a significant main effect of condition (p < 0.05), significant (p < 0.05) pairwise differences for each experimental condition versus the unloaded
499 condition are notated next to the condition Mean ± SD with an asterisk.

500 [^] clear difference versus the unloaded condition ([^] = possible, ^{^^} = likely, ^{^^^} = very likely).

501 [†] trivial difference versus the unloaded condition ([†] = possibly, ^{††} = likely, ^{†††} = very likely).

502 Percentage likelihoods are presented to the nearest whole number and thus the negative | trivial | positive effects may not always add up to 100%.

503 Table 4. Comparison of peak biceps femoris long head (BF_{lh}) muscle tendon unit length (as a % of resting length) and time of peak length (as a % of stride
 504 cycle) for all three conditions.

	Main effect (p)	Unloaded Mean ± SD	Thigh Mean ± SD	Shank Mean ± SD	Thigh-loaded versus unloaded		Shank-loaded versus unloaded			
					ES ± 95% CI	Percentage likelihood of		ES ± 95% CI	Percentage likelihood of	
						negative trivial positive effect	negative trivial positive effect			
Peak BF _{lh} length (%)	0.392	111.1 ± 1.2	111.4 ± 1.3	111.5 ± 1.1	0.27 ± 0.60	5 34 61	0.31 ± 0.32 ^{^^}	0 22 78		
Time occurrence of peak BF _{lh} length (%)	0.132	89.8 ± 3.1	90.4 ± 1.8	91.9 ± 2.5	0.22 ± 0.89	14 33 52	0.74 ± 1.08	4 9 87		

505 SD = standard deviation; ES = effect size (Cohen's d); CI = compatibility interval.
 506 Where there was a significant main effect of condition (p < 0.05), significant (p < 0.05) pairwise differences for each experimental condition versus the unloaded
 507 condition are notated next to the condition Mean ± SD with an asterisk.
 508 [^] clear difference versus the unloaded condition ([^] = possible, ^{^^} = likely, ^{^^^} = very likely).
 509 [†] trivial difference versus the unloaded condition ([†] = possibly, ^{††} = likely, ^{†††} = very likely).
 510 Percentage likelihoods are presented to the nearest whole number and thus the negative | trivial | positive effects may not always add up to 100%.