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

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Abstract

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

Introduction

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

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

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

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

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

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

Materials and methods

Participants

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

Data collection

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

An optical measurement system with infra-red light barriers (Optojump, Microgate, Italy) was placed either side of the sprint lane between 30 and 40 m to obtain spatiotemporal characteristics. A digital video camera (PXW-Z150, Sony, Japan) was set up perpendicular to the sprint lane at the 35 m mark (sampling frequency = 120 Hz, shutter speed = 1/725 s, resolution = 1920 × 1080 pixels). The camera was positioned 16 m from the centre of the lane, viewing the left side of the participants within a field of view approximately 10 m wide to ensure that one complete stride cycle of the left leg was captured for all trials. An 8 × 2 m plane was calibrated in the centre of the lane between the 31 and 39 m marks.

Determination of participant-specific wearable resistance loads

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

To ensure relevance to applied practice, it was decided that all participants would have 0.2 kg added to each shank during the shank-loaded condition. Using the parallel axis theorem, the mean knee angle (108.3°) determined from Zhong et al. (2017), and the segmental inertia parameters of de Leva (1996), a mass of 0.6 kg on each thigh was initially determined as providing comparable moment of inertia demands about the hip joint (approximately +4.5%) between both loaded conditions when averaged across the stride cycle, and that this would be achievable for all participants in the current study given their height and body mass. The directly measured thigh and shank lengths of each participant were then used along with each participant's mass and the segmental inertia parameters of de Leva (1996) to determine the participant-specific percentage distances (to the nearest whole percent) along each of the thigh and shank segments at which to place the centre of mass of the added loads in order to yield these increases in moment of inertia. Thigh loads were placed anteriorly at 76 ± 4% (range = 69 to 80%) of the distance from the proximal end of the segment across the studied group, and shank loads were placed anteriorly at 62 ± 11% (range = 42 to 78%) of the distance from the proximal end. This yielded increases of 4.48 ± 0.03% in the moment of inertia of the leg about the hip joint in the thigh-loaded condition, and 4.49 ± 0.01% in the shank-loaded condition (at knee angles of 108.3°). The rotational demands about the hip joint were therefore considered matched across an entire stride cycle between the two experimental conditions, with both ~4.5% higher than in the unloaded condition.

Data analysis

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

The raw video files were calibrated, after which a 6-point model (neck (mid C7-larynx), left hip, knee and ankle joint centres, left calcaneus and left 5th metatarsal) was manually digitised. Each trial was digitised twice from 10 frames prior to the initial touchdown until 10 frames after the next touchdown. All raw co-ordinate time-histories were then exported for subsequent analysis in Excel (Microsoft, USA) and Matlab. The mean co-ordinates across both digitisations were calculated and used for all subsequent analyses to reduce the potential random error associated with manual digitisation. Hip, knee and ankle joint angles were calculated at each frame, and MTU lengths of the biceps femoris long head (BFlh; chosen due to hamstring strain injuries primarily affecting this muscle; Koulouris & Connell, 2003; Askling, Tengvar, Saartok, & Thorstensson, 2007) were determined from the hip and knee flexion angles using the regression equations of Hawkins and Hull (1990). The raw joint angles and MTU lengths were low-pass filtered using a 4^{th} order Butterworth filter at 10 Hz, after which the padding frames were removed. All variables were then sampled at 101 evenly-spaced data points using an interpolating cubic spline to represent each 1% of the stride cycle from the first left foot touchdown (0%) to the next left foot touchdown (100%). The percentage of the stride cycle at which specific events (maximum knee flexion during ground contact, toe-off, maximum knee flexion during swing, maximum hip flexion during swing) occurred were then identified. Discrete values of the joint angle and MTU data at each event were extracted, and data were averaged across the two trials in each condition for each participant to obtain the dependent variables used for statistical analysis.

Statistical analysis

Given the intended high specificity of the loaded conditions to unloaded sprinting and the fact that previous studies with 2% BM increases on the thigh have observed moderate but non-significant effects on spatiotemporal variables during sprinting (Macadam et al., 2019), both traditional statistics and a magnitude-based decision approach (Batterham & Hopkins, 2006: Hopkins, 2019) were used to compare each of the loaded conditions with unloaded sprinting. Firstly, a repeated measures ANOVA (SPSS v. 26, IBM, USA) was conducted to determine if there was a significant ($p < 0.05$) main effect of condition. In such instances, pairwise comparisons were then conducted between each of the loaded conditions and the unloaded condition using Fisher's LSD. Secondly, effect sizes (*d*; Cohen, 1988) and their 95% compatibility intervals (Hopkins, 2019) were calculated between each of the loaded conditions and the unloaded condition. Thresholds of 0.2, 0.5 and 0.8 were used to define small, moderate and large mean effects, respectively (Cohen, 1988). Based on a smallest worthwhile effect size of 0.2 (Hopkins, 2004; Winter, Abt, & Nevill, 2014), clear effects were identified where the 95% compatibility interval did not overlap an effect size of both +0.2 and -0.2. The percentage likelihoods of a negative | trivial | positive effect were also calculated (Batterham & Hopkins, 2006). For clarity, any significant main effects are explicitly reported in the written text and any clear effects are described in the written results by reporting the magnitude threshold (e.g. small, moderate, large). The likelihoods of the clear effects (including the qualitative descriptors) are presented in the tables.

Results

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

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

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

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

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

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

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

Discussion and implications

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

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

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

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

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

There were no clear effects of either loading condition on knee angle at any of the three events studied during the stance phase (Table 3). Although these effects were unclear, the direction of the mean differences in the shank-loaded condition opposed those which occurred in the thigh-loaded condition (i.e. more flexed knee at touchdown and more extended knee at toe-off). Based on the relatively greater moment of inertia effects in the shank-loaded condition during late-swing and stance as discussed above, shank loading could lead to different effects during the stance phase. Zhang et al. (2019) observed a reduction in knee flexion at touchdown with ~0.37 kg loading per shank, but the lighter shank loads purposefully used in the current study may not have been sufficient to yield clear effects. During the swing phase, there was a possible reduction in maximum knee flexion in the thigh-loaded condition, but the mean effect size was less than small (d = 0.19). This may again be due to the relatively greater moment of inertia about the hip in the thigh-loaded than shank-loaded condition during the swing phase, but the reason why this might have affected knee flexion is not clear. One possible explanation is that this is a function of the thigh loading inhibiting hip flexion during swing (as discussed above), and thus the thigh segment reaches a less horizontal orientation which has a consequent effect on the knee angle between this and the shank.

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

Whilst one limitation of our study is that we did not compare the direct effect of different placements of the same absolute load, our study developed and described a novel objective method for matching the rotational demands about the hip joint between different loading configurations. This enabled us to assess the effects of two loading schemes which were theoretically matched for the overall rotational demands across an entire stride cycle, rather than observing likely increased effects with shank loading if matching the masses applied. Researchers should carefully consider the design of the loading protocols (e.g. matched total mass when greater shank overload is intended versus lower shank masses when matched rotational demands are intended) depending on their specific question. Further limitations relate to the use of two-dimensional motion analysis, as well as an optical measurement system for determining step characteristics which likely led to a small over-estimation in contact time and an under-estimation in flight time compared with previous research which has used force platforms (e.g. Macadam et al., 2019), but these effects were consistent across all studied conditions and thus do not limit our comparisons. Finally, our results are also from a relatively small sample of university-level sprinters and further investigations are required to assess the generalisability of these findings. Future research should also consider the acute neuromuscular, physiological and endocrine responses to training with light wearable resistance so that such sessions can be best programmed. This will also help to inform the planning of longer-term training interventions which are ultimately required to assess whether training with light wearable resistance can enhance sprinting performance.

Conclusion

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

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Declaration of interest statement

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

References

Alcaraz, P. E., Palao, J. M., Elvira, J. L., & Linthorne, N. P. (2008). Effects of three types of resisted sprint training devices on the kinematics of sprinting at maximum velocity. *Journal of Strength and Conditioning Research, 22*(3), 890-897.

Askling, C. M., Tengvar, M., Saartok, T., & Thorstensson, A. (2007). Acute first-time hamstring strains during high-speed running: a longitudinal study including clinical and magnetic resonance imaging findings. *American Journal of Sports Medicine*, *35*(2), 197-206.

Batterham, A. M., & Hopkins, W. G. (2006). Making meaningful inferences about magnitudes. *International Journal of Sports Physiology and Performance*, *1*(1), 50–57.

Bennett, J. P., Sayers, M. G., & Burkett, B. J. (2009). The impact of lower extremity mass and inertia manipulation on sprint kinematics. *Journal of Strength and Conditioning Research*, *23*(9), 2542–2547.

Bompa, T. O. (1999). *Periodization: theory and methodology of training* (4th edn). Champaign, IL: Human Kinetics.

Bondarchuk, A. P. (2007). *Transfer of training in sports*. Muskegon, MI: Ultimate Athlete Concepts.

Clark, K. P., Stearne, D. J., Walts, C. T., & Miller, A. D. (2010). The longitudinal effects of resisted sprint training using weighted sleds vs. weighted vests. *Journal of Strength and Conditioning Research*, *24*(12), 3287-3295.

Cohen, J. (1988). *Statistical power analysis for the behavioural sciences* (2nd edn). New York City, NY: Lawrence Erlbaum Associates.

Cronin, J., Ogden, T., Lawton, T., & Brughelli, M. (2007). Does increasing maximal strength improve sprint running performance? *Strength and Conditioning Journal*, *29*(3), 86–95.

Cronin, J., Hansen, K., Kawamori, N., & McNair, P. (2008). Effects of weighted vests and sled towing on sprint kinematics. *Sports Biomechanics*, *7*(2), 160-172.

de Leva, P. (1996). Adjustments to Zatsiorsky-Seluyanov's segment inertia parameters. *Journal of Biomechanics*, *29*(9), 1223-1230.

Delecluse, C. (1997). Influence of strength training on sprint running performance: current findings and implications for training. *Sports Medicine*, *24*(3), 147-156.

D'Souza, D. (1994). Track and field athletics injuries - a one-year survey. *British Journal of Sports Medicine*, *28*(3), 197–202.

Haugen, T., Danielsen, J., Alnes, L. O., McGhie, D., Sandbakk, Ø., & Ettema, G. (2018). On the importance of "front-side mechanics" in athletics sprinting. *International Journal of Sports Physiology and Performance*, *13*(4), 420-427.

Hawkins, D., & Hull, M. L. (1990). A method for determining lower-extremity muscle tendon lengths during flexion extension movements. *Journal of Biomechanics*, *23*(5), 487–494.

Hopkins, W. G. (2004). How to interpret changes in an athletic performance test. *Sportscience*, *8*, 1–7.

Hopkins, W. G. (2019). Magnitude-based decisions. *Sportscience*, *23*, i-iii.

Kenneally-Dabrowski, C. J. B., Brown, N. A. T., Lai, A. K. M., Perriman, D., Spratford, W., & Serpell, B. G. (2019). Late swing or early stance? A narrative review of hamstring injury mechanisms during high-speed running. *Scandinavian Journal of Medicine and Science in Sports*, *29*(8), 1083-1091.

Koulouris, G., & Connell, D. (2003). Evaluation of the hamstring muscle complex following acute injury. *Skeletal Radiology*, *32*(10), 582-589.

Macadam, P., Simperingham, K. D., & Cronin, J. B. (2017a). Acute kinematic and kinetic adaptations to wearable resistance during sprint acceleration. *Journal of Strength and Conditioning Research*, *31*(5), 1297-1304.

Macadam, P., Cronin, J. B., & Simperingham, K. D. (2017b). The effects of wearable resistance training on metabolic, kinematic and kinetic variables during walking, running, sprint running and jumping: a systematic review. *Sports Medicine*, *47*(5), 887-906.

Macadam, P., Nuell, S., Cronin, J. B., Uthoff, A. M., Nagahara, R., Neville, J., Graham, S. P., & Tinwala, F. (2019). Thigh positioned wearable resistance affects step frequency not step length during 50 m sprint-running. *European Journal of Sport Science*, ahead of print.

Pajić, Z., Kostovski, Ž., Ilić, J., Jakovljević, S., & Preljević, A. (2011). The influence of inertial load application on kinematic and dynamic performances of running at maximum speed phase. *Sport Science*, *4*(1), 107-112.

Ropret, R., Kukolj, M., Ugarkovic, D., Matavulj, D., & Jaric, S. (1998). Effects of arm and leg loading on sprint performance. *European Journal of Applied Physiology and Occupational Physiology*, *77*(6), 547–550.

Simperingham, K., & Cronin, J. (2014). Changes in sprint kinematics and kinetics with upper body loading and lower body loading using Exogen™ exoskeletons: A pilot study. *Journal of Australian Strength and Conditioning*, *22*(5), 69–72.

Smith, G. (1989). Padding point extrapolation techniques for the butterworth digital filter. *Journal of Biomechanics*, *22*(8-9), 967–971.

Spinks, C. D., Murphy, A. J., Spinks, W. L., & Lockie, R. G. (2007). The effects of resisted sprint training on acceleration performance and kinematics in soccer, rugby union, and Australian football players. *Journal of Strength and Conditioning Research*, *21*(1), 77-85.

Wild, J., Bezodis, N. E., Blagrove, R., & Bezodis, I. (2011). A biomechanical comparison of accelerative and maximum velocity sprinting: specific strength training considerations. *Professional Strength & Conditioning*, *21*, 23-36.

Winter, E. M., Abt, G. A., & Nevill, A. M. (2014). Metrics of meaningfulness as opposed to sleights of significance. *Journal of Sports Sciences, 32*(10), 901–902.

Yeung, S.S., Suen, A. M., & Yeung, E. W. (2009). A prospective cohort study of hamstring injuries in competitive sprinters: preseason muscle imbalance as a possible risk factor. *British Journal of Sports Medicine*, *43*(8), 589-594.

Young, W. B. (2006). Transfer of strength and power training to sports performance. *International Journal of Sports Physiology and Performance*, *1*(2), 74-83.

Zhang, C.,Yu, B., Yang, C., Yu, J., Sun, Y., Wang, D., Yin, K., Zhuang, W., & Liu, Y. (2019). Effects of shank mass manipulation on sprinting techniques. *Sports Biomechanics*, doi: 10.1080/14763141.2019.1646796.

Zhong, Y., Fu, W., Wei, S., Li, Q., & Liu, Y. (2017). Joint torque and mechanical power of lower extremity and its relevance to hamstring strain during sprint running. *Journal of Healthcare Engineering*, *2017*, 8927415.

Tables

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

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

SD = standard deviation; ES = effect size (Cohen's d); CI = compatibility interval.
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 condition are notated next to the condition Mean ± SD with an asterisk.
^ clear difference versus the unloaded condition (^ = possible, ^^ = likely. ^^^ = very likely).
† trivial difference versus the unloaded condition († = possibly, †† = likely, ††† = very likely).
Percentage likelihoods are presented to the nearest whole number and thus the negative | trivial | positive effects may not always add up to 100%.

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

					Thigh-loaded versus unloaded		Shank-loaded versus unloaded	
Event	Main effect (p)	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 (°)	0.751	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

SD = standard deviation; ES = effect size (Cohen's d); CI = compatibility interval.
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 condition are notated next to the condition Mean ± SD with an asterisk.
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Percentage likelihoods are presented to the nearest whole number and thus the negative | trivial | positive effects may not always add up to 100%.

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

					Thigh-loaded versus unloaded		Shank-loaded versus unloaded	
Event	Main effect (p)	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

SD = standard deviation; ES = effect size (Cohen's d); CI = compatibility interval.
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 condition are notated next to the condition Mean ± SD with an asterisk.
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Percentage likelihoods are presented to the nearest whole number and thus the negative | trivial | positive effects may not always add up to 100%.

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

					Thigh-loaded versus unloaded		Shank-loaded versus unloaded	
	Main effect (p)	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
Peak BFlh 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 BFlh 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

SD = standard deviation; ES = effect size (Cohen's d); CI = compatibility interval.
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 condition are notated next to the condition Mean ± SD with an asterisk.
^ clear difference versus the unloaded condition (^ = possible, ^^ = likely. ^^^ = very likely).
† trivial difference versus the unloaded condition († = possibly, †† = likely, ††† = very likely).
Percentage likelihoods are presented to the nearest whole number and thus the negative | trivial | positive effects may not always add up to 100%.