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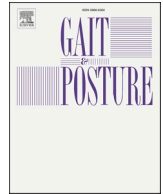
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Lower limb EMG activation during reduced gravity running on an incline. Speed matters more than hills irrespective of indicated bodyweight

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ABSTRACT

Background: Progressive loading of the lower limb muscles during running on a positive pressure or reduced gravity (Alter-G™) treadmill is suggested as a rehabilitation strategy after muscle and tendon injury but the influence of running up or downhill and at higher speeds is not known, nor are the interaction effects of speed, inclination, and indicated bodyweight.

Research question: What are the lower limb EMG activation levels and cadence when running up and downhill in normal and reduced gravity?

Methods: 10 recreationally active male athletes ran on a positive-pressure Alter-G™ treadmill at: 3 indicated bodyweights (60 %, 80 %, and 100 %); 5 speeds (12, 15, 18, 21, and 24 km/h); for incline, decline, and flat conditions (-15 %, -10 %, -5%, 0%, 5%, 10 %, and 15 %); while monitoring the surface EMG of 11 leg muscles as well as cadence (strides per minute).

Results and significance: Linear mixed models showed significant effect of running speed, inclination, and indicated bodyweight, with interaction effects observed. Increasing running speed was associated with the largest change in activity, with smaller effects for increasing bodyweight and inclination. Downhill running was associated with reduced activity in all muscle groups, and more tightly clustered activity patterns independent of speed. Substantial variation in sEMG activity occurred in the flat and uphill conditions. Subject responses were quite variable for sEMG, less so for cadence. For the conditions examined, increasing running speed induced the largest changes in EMG of all muscles examined with smaller changes seen for manipulations of inclination and bodyweight.

1. Introduction

Progressively increased loading is considered a mainstay of rehabilitation and training after muscle and tendon injury. Quantifying the magnitudes of the loads applied during training is therefore crucial to appropriately prescribing training and rehabilitation programmes. For a given duration, runners can manipulate: running speed, inclination, and recently also their indicated bodyweight as a means of varying the “on legs” loading. Reduced gravity treadmills such as the Alter-G™, have users wearing neoprene shorts that are sealed (zipped) into a chamber surrounding the treadmill. The chamber then inflates, and the increasingly positive air pressure exerted on the shorts ‘unloads’ the user’s bodyweight being borne through the deck. By calibrating the air pressure against the measured bodyweight the system then allows ‘reduced

gravity’ walking and running. Typically we notice clinicians during rehabilitation using varying bodyweight reduction from no reduction (i.e. 100 % of bodyweight borne through the deck) down to perhaps a 60 % reduction (i.e. only 40 % of the user’s bodyweight is borne through the deck). When rehabilitating an athlete from a muscle injury, it is assumed that reducing indicated bodyweight is associated with reductions in muscle loads [1] however it is not known how these changes influence muscle activity, nor how manipulating indicated bodyweight interacts with running speed and/or inclination changes. Muscle and tendon loads are impossible to measure non-invasively in vivo. Electromyography is often used as a proxy for muscle load during exercise, although this approach has limitations [2,3]. Relative changes in surface EMG excitation (amplitude) for the same muscle performing similar activities during the same session are suggested to be valid methods of comparing

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relative muscle activity and therefore muscle force production for these different conditions, with some limitations [3]. During cyclic activity such as gait muscle work can be examined as a “peak” activation (during the cycle), or a sum of the entire activation during this cycle. Both approaches have merit. In a clinical situation where healing muscle is being considered it may be appropriate to examine the peak muscle activation associated with a certain exercise and its parameters which could represent a “worst case” in terms of muscle load to allow safe planning of loading increments.

As there are no guidelines or even any data available for muscle activation during clinically applied parameters of Alter-*G*TM training, practitioners are forced to estimate the effects of changing these parameters on muscle and tendon loading which may potentially over- or underload the muscle or tendon of interest. Specifically where a clinician has the option of changing running speed, inclination, and percent indicated bodyweight, they are forced to estimate the relative activation levels for different muscles at these different possible combinations. If the relative changes are very small, (e.g. a few percent increase or decrease) there is likely little or no clinical effect of such manipulations. If these changes are large (e.g. doubling or more the relative muscle activation) then such manipulations could have a dramatic effect on tissue loading during rehabilitation [4].

Previous research using positive pressure treadmills have examined plantar pressures [1] at slower speeds including walking [5,6]. However during athletic rehabilitation larger loads are seen with higher running speeds which have not been well explored. Very little information on muscle activity is available regarding downhill running [7] and to our knowledge there is no information on activation patterns during downhill running with reduced bodyweight at higher speeds.

Accordingly, this study sought to investigate the effect of manipulating running speeds, inclination, and indicated bodyweight on peak EMG measures of 11 lower limb muscles. Additionally we report the associated cadence changes.

2. Methods

Ten healthy, trained male athletes volunteered to take part in this study (age: 28 ± 5 yrs, body mass: 73 ± 8 kg, height 180 ± 6 cm). Participants self-reported that they regularly ran at speeds greater than 24 km/h and in the 3 months leading into the study ran 31 ± 15 km, on average, per week. After being fitted with appropriately sized neoprene Alter-*G* shorts, all subjects then familiarised and warmed up initially with a 5-minute walk at 5 km/h, then 3 min run at 10 km/h, and finally, when they indicated they felt ready, two 10 s efforts each at 21 and 24 km/h respectively. The testing then commenced which comprised a block-randomized sequence of 78 combinations of 5 running speeds (12–24 km/h), 3 indicated bodyweights (60 %, 80 %, and 100 %), and 7 inclinations (–15 % to +15 %). (Note that when running in reverse for the downhill conditions, the Alter-*G*TM treadmill has a maximum speed of 15 km/h, hence only 2 downhill running speed conditions were examined).

Informed consent was sought and obtained for each subject prior to data collection. Local ethics approval was obtained (Anti-Doping Lab Qatar Approval number E2018000272).

Collection of surface electromyography (sEMG) data was performed with wireless sEMG (Delsys Trigno, Boston, MA) using an acquisition frequency of 2000 Hz. Before electrode placement, the subject's skin was shaved and then cleaned with alcohol, in accordance with the SENIAM (Surface ElectroMyoGraphy for the NonInvasive Assessment of Muscles) guidelines. Rectangular electrodes measuring $37 \times 26 \times 15$ mm (Delsys Trigno, Boston, MA) were placed on the right leg of the following 11 muscles: semitendinosus, biceps femoris, vastus lateralis, vastus medialis, rectus femoris, gastrocnemius lateralis, gastrocnemius medialis, soleus, gluteus maximus, gluteus medius, and tibialis anterior. An additional sensor was placed on the mid-tibia, with its vertical accelerometer trace used to identify heel strike [5]. For each trial condition

participants were instructed to run until they felt comfortable with the given configuration of speed, inclination, and bodyweight, and then indicate the point where their gait felt “normal”. At this point EMG collection began and continued for 30 s. Subsequent analysis included a minimum of 20 strides for each condition, for each subject. These data were initially analysed using custom MATLAB scripts allowing visualisation and analyses of each stride of each trial for each participant. Initially the data were rectified, and then filtered with a 2nd order Butterworth filter with a 30 Hz cutoff frequency, before being normalized to the maximum value seen across any trial for the given subject and muscle [8]. Each stride was broken into 101 equal sections (0–100 %, right heel strike to right heel strike) for further analysis. Since the EMG traces of the individual strides were not normally distributed, the median stride was taken as representative for each subject, in each condition. (Supplementary Fig. 12 shows representative samples of the EMG traces for each of the 11 muscles for a selected subject in a single condition.) These data were then exported to JMP Pro 14 for further analysis. Initially distributions were examined using frequency histograms, Q-Q plots, and Shapiro Wilk testing. Descriptive statistics were then calculated, and subsequently linear mixed models with fixed effects of: speed, inclination, and bodyweight, and random (subject) effects were then used to identify main differences and interaction effects. Additionally cadence was calculated for each subject, in each trial condition, as the average number of strides (right foot strikes) per minute [9]. These data were examined similarly - initially distributions and normality and then linear mixed models to identify main and interaction effects.

3. Results

Table 1 shows the F Ratio and associated p value for each main effect and interaction considered for each muscle for the peak EMG activation (total integrated, iEMG, is shown in supplementary Table 1) as well as cadence. Due to the large amount of data generated and the similar findings for peak and iEMG, subsequent analyses focus on peak EMG only. These data are summarised in the figures showing individual data along with lines of best fit with their associated confidence intervals for the 3 bodyweight conditions. Fig. 1 shows the rectus femoris, vastus lateralis and medialis average peak activation. Fig. 2 shows the posterior thigh muscles: gluteus maximus, semitendinosus, and biceps femoris. Fig. 3 shows the calf muscles: soleus, medial and lateral gastrocnemius, and finally Fig. 4 shows the gluteus medius and tibialis anterior muscles. Generally it can be seen that for all muscles the peak activation increased with increasing running speed, independent of inclination and indicated bodyweight. The changes associated with the different bodyweights were not as clear (note the overlap of the 95 % confidence intervals of the polynomial regression lines of best fit). The downhill conditions were seen to have slightly lower activation levels for most muscles and speeds, however these effects too were relatively smaller than those seen for changing running speed. In contrast the effect of bodyweight change on cadence was clearer (Fig. 5). While the strong effect of increasing cadence with increasing running speed is present, a clearer effect of increased cadence with increased indicated bodyweight can be appreciated through the smaller overlap of the 95 % confidence intervals at all inclinations and running speeds. Supplementary Figs. 1–11 interactively show muscle activation as a function of running speed, inclination, and bodyweight as 3-dimensional scatter plots. Representative EMG traces for all muscles for a single subject, in a single trial condition are provided in Supplementary Fig. 12.

4. Discussion

For all muscles examined, the largest changes in muscle activity were seen for the alterations in running speed whereas smaller changes were seen for manipulations of inclination and bodyweight (Figs. 1–4, Supplementary Figs. 1–11, Fig. 6). Lower activation was seen almost

Table 1

F Ratios and p-values for each main and interaction effects for each of the muscles examined for peak EMG, and cadence (strides/minute). “Speed” is the 5 running speed conditions (12, 15, 18, 21, and 24 km/h), “Incline” is 7 conditions (-15 %, -10 %, -5 %, 0 %, 5 %, 10 %, and 15 %), and there were 3 “Bodyweight” conditions (60 %, 80 %, and 100 % indicated bodyweight). Statistically significant results are highlighted in boldface with an asterisk (*).

Peak EMG	Semitendinosus		Biceps Femoris		Rectus Femoris		Vastus Lateralis		Vastus Medialis	
	F Ratio	Prob > F	F Ratio	Prob > F	F Ratio	Prob > F	F Ratio	Prob > F	F Ratio	Prob > F
Speed	294.200	<.0001*	119.102	<.0001*	493.184	<.0001*	151.28	<.0001*	190.404	<.0001*
Incline	15.004	0.0001*	19.202	<.0001*	33.844	<.0001*	22.038	<.0001*	23.006	<.0001*
Bodyweight	35.835	<.0001*	20.451	<.0001*	24.037	<.0001*	127.204	<.0001*	165.555	<.0001*
Speed*Incline	5.408	0.0203*	9.909	0.0017*	19.302	<.0001*	11.197	0.0009*	0.052	0.8194
Speed*Bodyweight	3.460	0.0632	0.736	0.3911	4.087	0.0435*	2.276	0.1318	1.776	0.183
Incline*Bodyweight	2.619	0.106	1.685	0.1946	3.475	0.0627	7.973	0.0049*	0.121	0.7281
Speed*Incline*Bodyweight	0.662	0.4161	0.404	0.5252	5.923	0.0152*	2.674	0.1024	0.291	0.5899

Peak EMG (cont'd)	Gluteus Maximus		Tibialis Anterior		Soleus		Gastrocnemius Lateralis		Gastrocnemius Medialis		Gluteus Medius	
	F Ratio	Prob > F	F Ratio	Prob > F	F Ratio	Prob > F	F Ratio	Prob > F	F Ratio	Prob > F	F Ratio	Prob > F
Speed	150.003	<.0001*	209.521	<.0001*	104.602	<.0001*	102.816	<.0001*	64.152	<.0001*	97.145	<.0001*
Incline	34.551	<.0001*	0.483	0.4871	16.479	<.0001*	11.956	0.0006*	4.283	0.0388*	3.274	0.0708
Bodyweight	16.255	<.0001*	31.565	<.0001*	20.157	<.0001*	10.458	0.0013*	5.098	0.0242*	6.927	0.0087*
Speed*Incline	5.190	0.0230*	0.211	0.6464	1.662	0.1977	0.901	0.3428	0.007	0.9322	0.860	0.354
Speed*Bodyweight	3.136	0.077	8.697	0.0033*	0.165	0.6848	0.538	0.4636	0.900	0.343	2.231	0.1357
Incline*Bodyweight	2.459	0.1173	0.426	0.5144	0.868	0.3517	0.600	0.439	0.856	0.3551	1.462	0.227
Speed*Incline*Bodyweight	0.312	0.5769	0.431	0.5118	0.568	0.4515	0.008	0.9298	0.067	0.7953	1.467	0.2262

Cadence (strides/min)		
	F Ratio	Prob > F
Speed	343.972	<.0001*
Incline	105.493	<.0001*
Bodyweight	82.932	<.0001*
Speed*Incline	29.553	0.0003*
Speed*Bodyweight	7.971	0.015*
Incline*Bodyweight	2.972	0.1054
Speed*Incline*Bodyweight	3.755	0.075

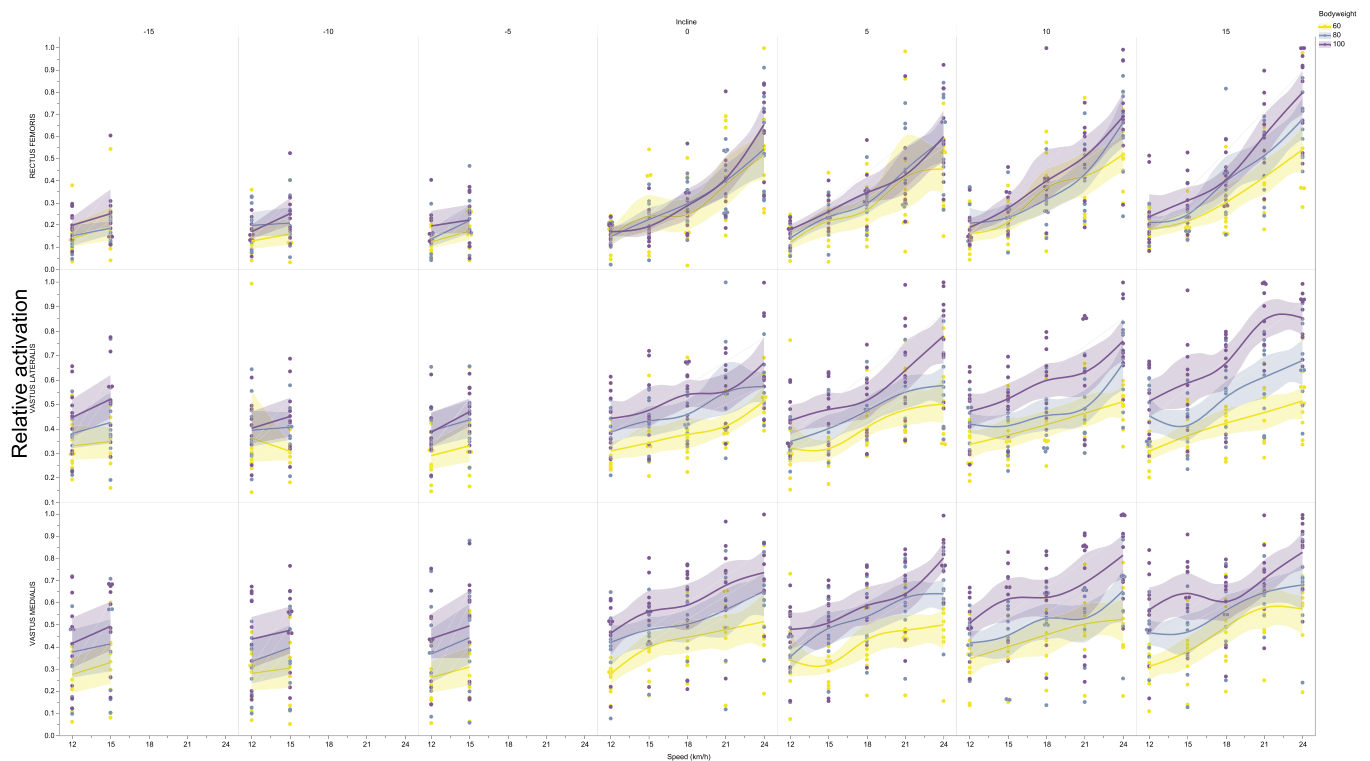


Fig. 1. Group estimators for the Rectus Femoris (upper panels) Vastus Lateralis (middle) and Vastus Medialis (lower) muscles examined at each of the speeds (lower x-axis for each panel) and inclinations (upper panel, from -15 % downhill to 15 % uphill, left to right). The fitted splines and confidence intervals are colour-coded representing the different bodyweight condition: purple (100 %), green (80 %) and yellow (60 %). Relative activation is from 0 to 1 for each panel. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

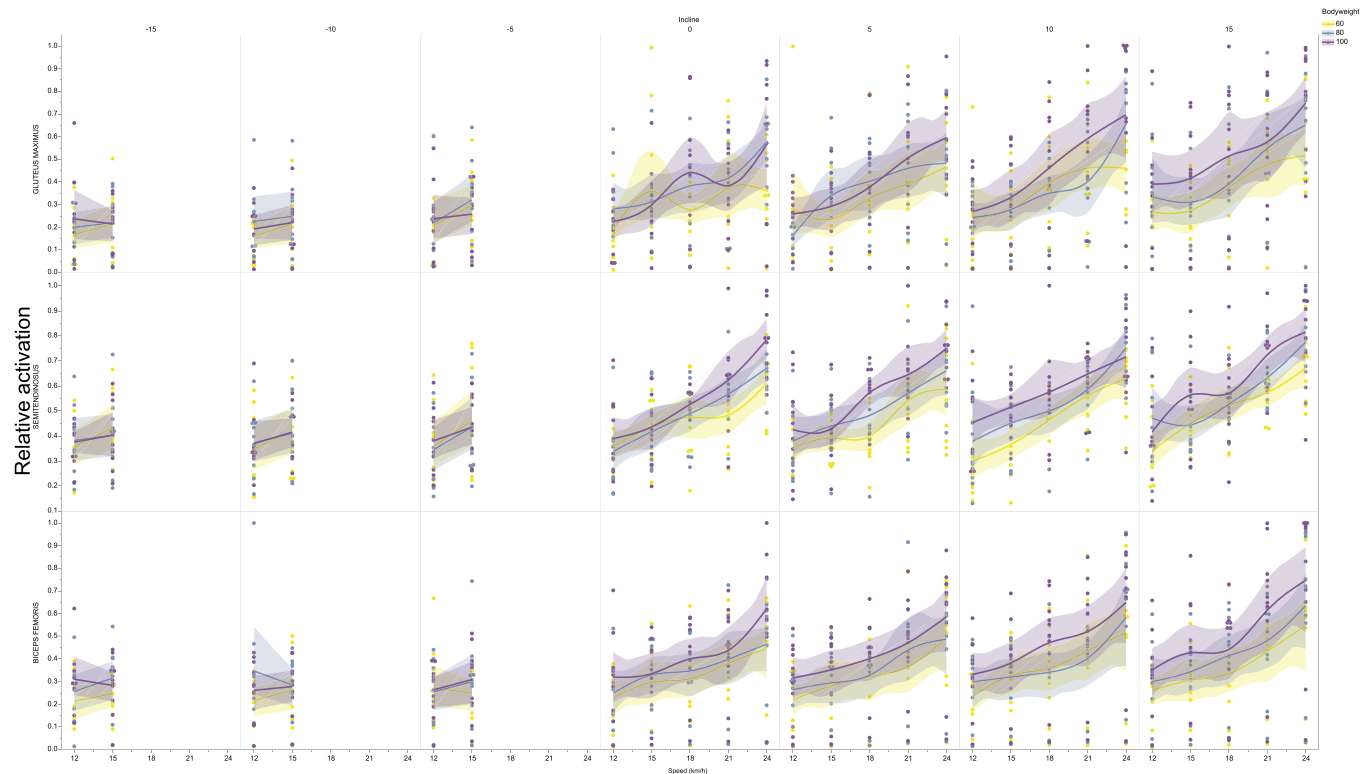


Fig. 2. Group estimators for the Gluteus maximus (upper panels) Semitendinosus (middle) and Biceps Femoris (lower) muscles examined at each of the speeds (lower x-axis for each panel) and inclinations (upper panel, from -15 % downhill to 15 % uphill, left to right). The fitted splines and confidence intervals are colour-coded representing the different bodyweight condition: purple (100 %), green (80 %) and yellow (60 %). Relative activation is from 0 to 1 for each panel. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

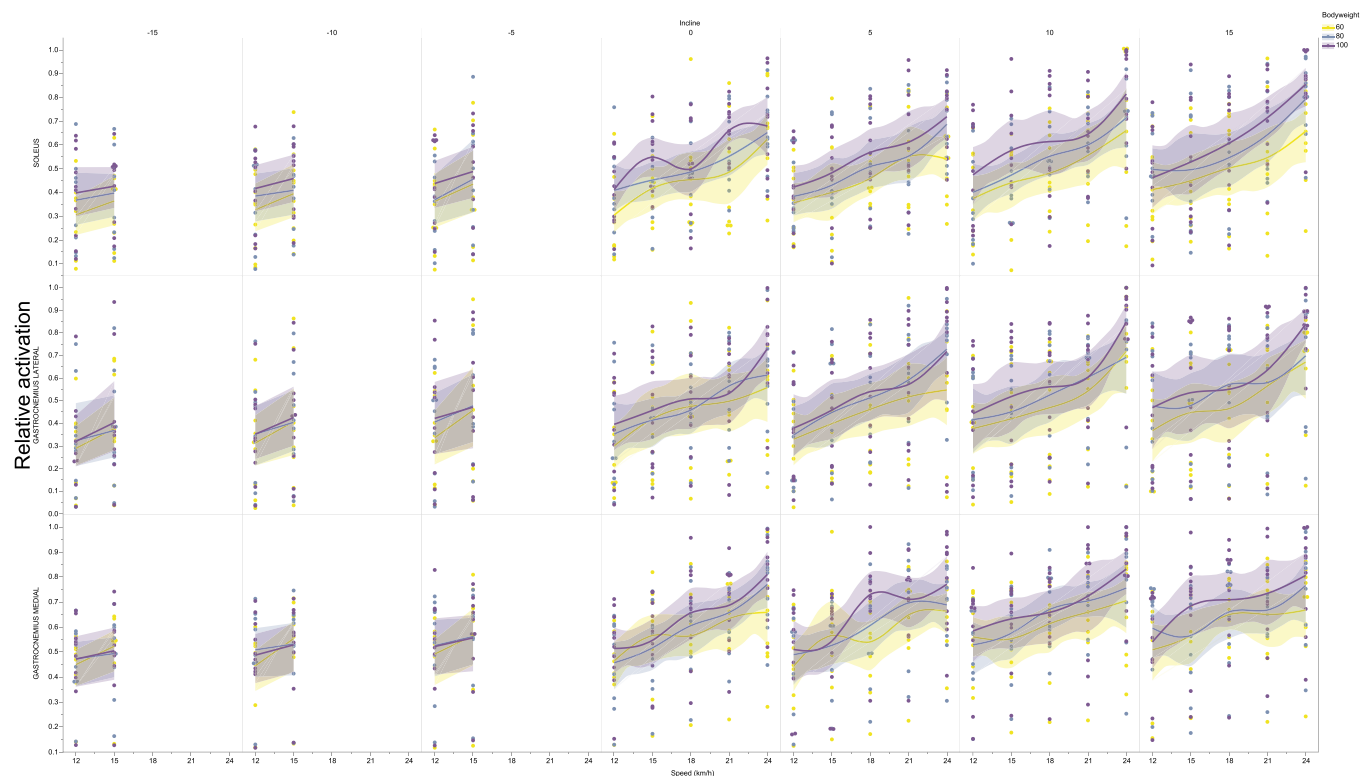


Fig. 3. Group estimators for the Soleus (upper panels) Lateral Gastrocnemius (middle) and Medial Gastrocnemius (lower) muscles examined at each of the speeds (lower x-axis for each panel) and inclinations (upper panel, from -15 % downhill to 15 % uphill, left to right). The fitted splines and confidence intervals are colour-coded representing the different bodyweight condition: purple (100 %), green (80 %) and yellow (60 %). Relative activation is from 0 to 1 for each panel. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

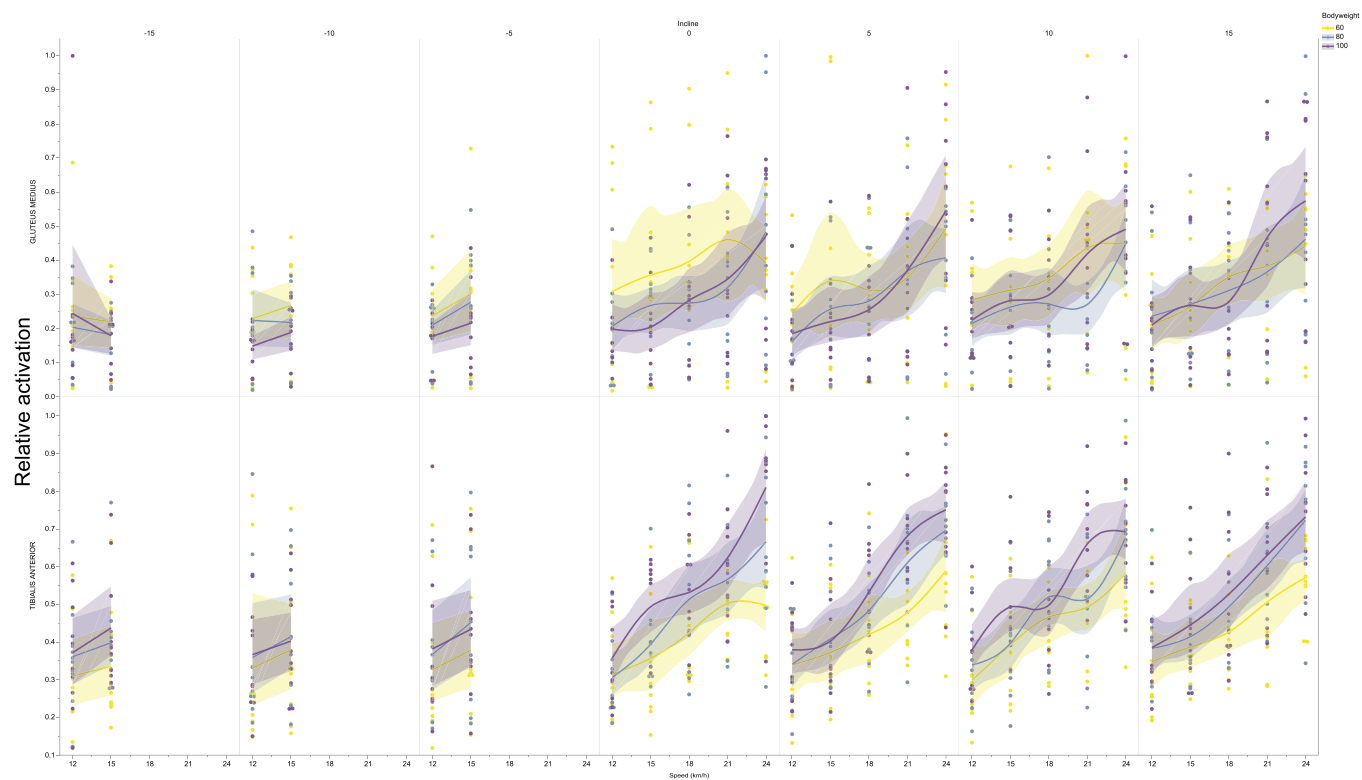


Fig. 4. Group estimators for the Gluteus medius (upper panels) and Tibialis Anterior (lower panels) muscles examined at each of the speeds (lower x-axis for each panel) and inclinations (upper panel, from -15 % downhill to 15 % uphill, left to right). The fitted splines and confidence intervals are colour-coded representing the different bodyweight condition: purple (100 %), green (80 %) and yellow (60 %). Relative activation is from 0 to 1 for each panel. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

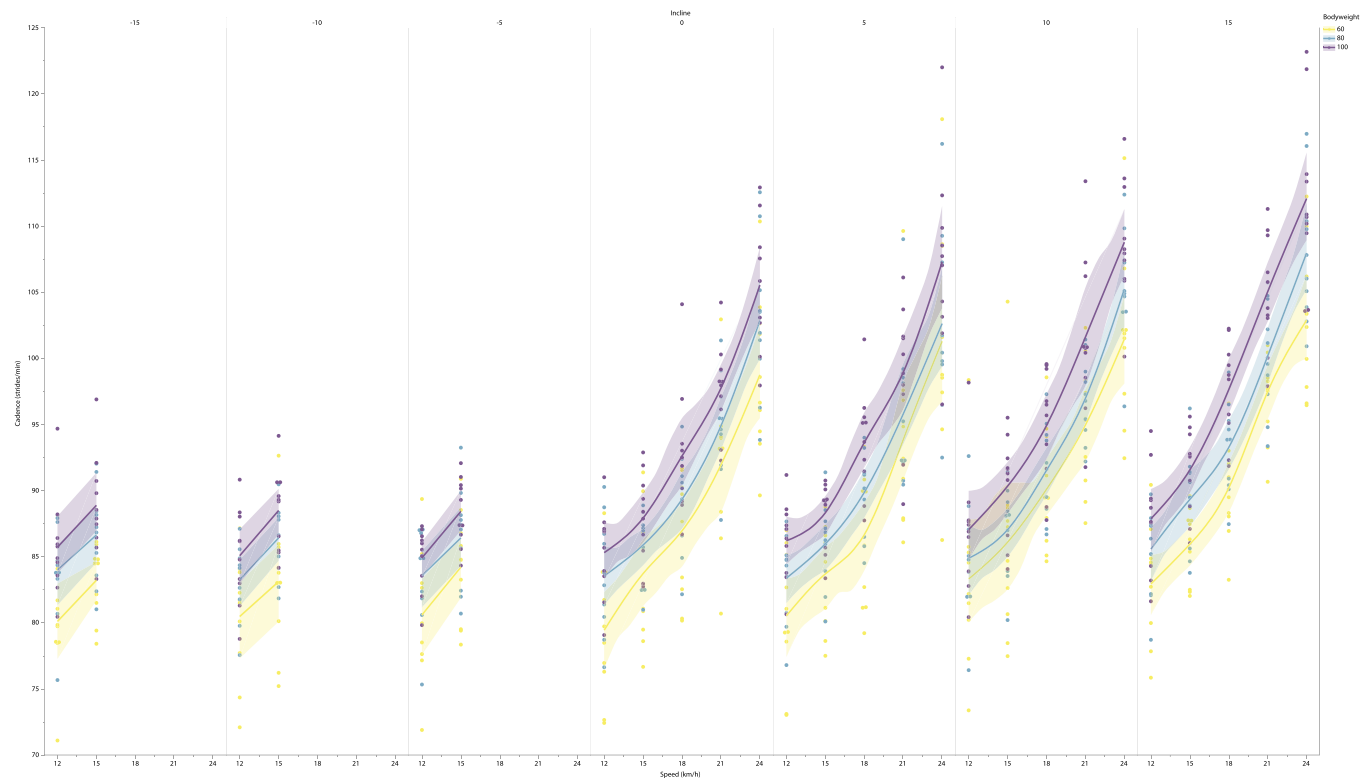


Fig. 5. Cadence (strides per minute) for each of the running speeds (lower horizontal axis) at each of the inclinations (upper horizontal axis) at each of the indicated bodyweights (purple: 100 %, green: 80 %, and yellow: 60 %). (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

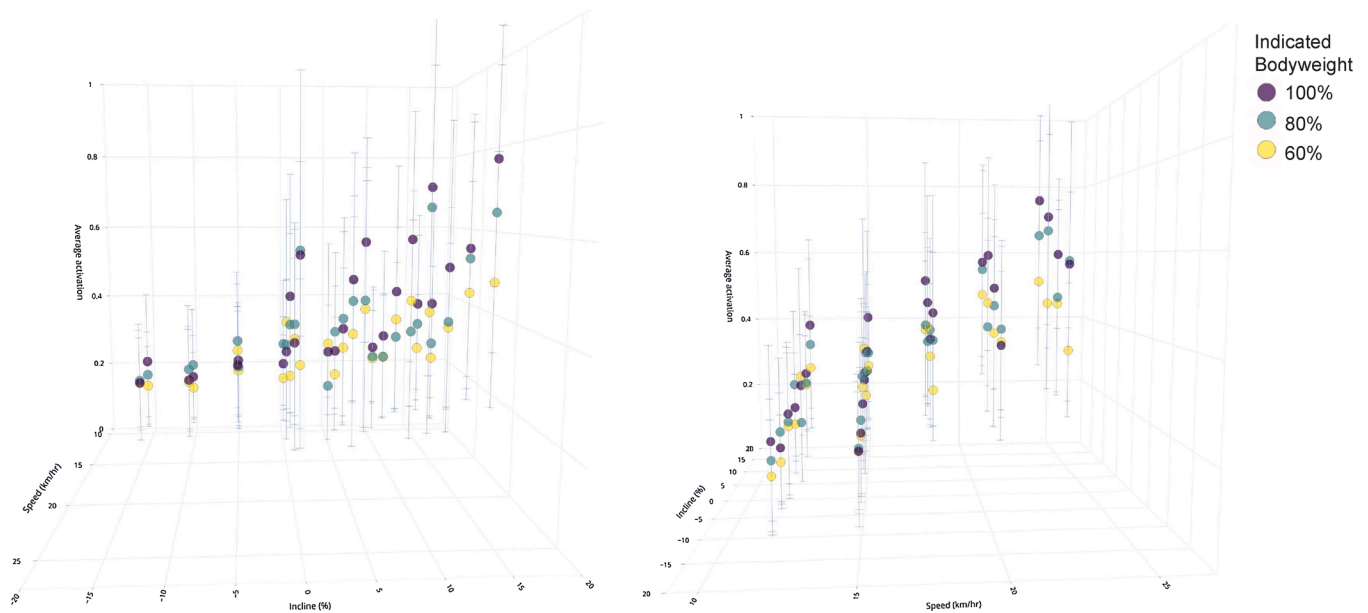


Fig. 6. Two views of an online supplementary data visualisation (<https://plotly.com/~rodw/23/>). This example shows 2 rotated views of the same 3-dimensional scatter chart. Vertical axes on both charts represent peak normalized EMG activation of the gluteus maximus muscle (0-1). Horizontal axes are running speed and incline, and the individual points are coloured according to the indicated bodyweight condition (purple:100 %, green: 80 %, yellow: 60 %). Whiskers represent standard deviations for each group mean. The online charts allow rotation, panning, and zooming for further exploration, and includes the raw data table. In this case the visualisations suggest that the (low) activation for gluteus maximus is relatively unchanged for each of the downhill conditions at each of the indicated bodyweights, however as the inclination increases from positive, a steeper increase in activation is evident (left image). In contrast the right image shows a relatively consistent effect of running speed and increasing activation. This pattern of more tightly clustered, lower activation for the downhill conditions, and increasing activation and spread of conditions was evident to different degrees across all muscles examined. (For interpretation of the references to colour in this figure legend, the reader is referred to the web version of this article).

uniformly across all the downhill conditions, with more tightly clustered low activation levels independent of the inclination, speed, and indicated bodyweight.

Clinically it may be inferred that changing running speed is by far the strongest driver of muscle activation in comparison to bodyweight or inclination. The present data show broad group trends and large inter-individual variation in response to the given loading conditions. While speed is the biggest driver of muscle activity change, the between subject variability is so large as to force caution when attempting to infer subject-specific results from these group data.

The non-linear nature of the relationship between higher running speeds and activation levels concurs with previous research [10] which documented a similar finding for work performed by the hamstring muscles during high speed running. The speeds examined here, while typically higher than those seen in other treadmill EMG studies [11] were not as high as those recorded by Schache et al. [10], and serve as a reminder that extrapolation from lower speed running and walking to very high speed running should be done with extreme care, if at all.

The exploratory and descriptive nature of this research is forced by the experimental design which had 78 conditions (and therefore 3081 possible pairwise comparisons) for each of the 11 muscles. Conducting all possible comparisons among these would lead to many false positive findings, therefore we have only presented the F-ratio and associated p-values, along with the data visualisations for the model (and their raw data). Subsequent hypothesis driven research may use these data to inform a priori power calculations for different experimental designs, but we are unwilling to try to present definitive pairwise comparisons. With this important limitation in mind, we cautiously present some observations we believe to be clinically relevant. We encourage interested readers to examine the online supplementary data visualisations and their associated data tables.

4.1. Downhill running has lower peak EMG for the quadriceps to a point

Using an inverse dynamics approach Park et al. [12] showed that downhill running at approximately 12 km/h (6% and 9% downhill) progressively increased the work done at the knee at the expense of the ankle in comparison to flat running. Increasing the downhill slope at this lower speed running was associated with an increase in knee range of motion and a reduction in hip and ankle range of motion. The current data complement this finding showing less effect of manipulating the inclination on EMG of the vasti. We speculate that the lower EMG seen in the current work can be reconciled to Park et al.'s [12] findings of increased work done in downhill running by recalling the likely eccentric nature of these muscles' action during descent which would be associated with a lower voltage for a given force output [13,14].

Vernillo et al. [11] suggested that previous research [12] had failed to show an increase in EMG of the vasti during downhill running as the inclinations were only 5% and needed to be more than 7% before such differences would appear. The 3 downhill conditions examined here span these ranges, and we cautiously suggest that there may be both a muscle-specific and non-linear effect. The current study is underpowered for such a post-hoc comparison, however we suggest that the interplay of relative unloading of the muscles at 5% decline then shifts in some muscles at the greater declines, especially 15 % where we postulate more braking force is required in landing. Visual inspection of the traces for vastus lateralis and medialis appear to show a drop in activity from 5% to 10 % downhill, but then an increase when running downhill at 15 % (Supplementary figures and online data). By contrast the rectus femoris (perhaps due to the eccentric hip flexion component during stance phase) shows a small, steady increase in activity from 5% to 15 % downhill.

4.2. Changing indicated bodyweight and inclination doesn't change peak EMG as much as you might think

Examining more than one running speed condition allowed us to see that these previously documented effects of bodyweight change were dwarfed by changing running speed. Hunter et al. [15] initially suggested reducing bodyweight as a clinical strategy to unload lower limb muscles during running in a study of 11 male cross-country athletes running at a fixed (16 km/h) speed. Similarly, Sainton et al. [16] considered only a single preferred running speed when describing the effect of muscle activity on different levels of bodyweight support. While the group effect of increasing indicated bodyweight in the present study was generally to show higher activation, the effect sizes seen here were not as large as seen for manipulation of running speed. Intuitively it may seem that nearly doubling the increased bodyweight from 60 % to 100 % should have a large effect on the activation of the antigravity muscles, however these effect sizes were much smaller than those seen through the changes in running speed seen across this experiment. Altering the inclination of the treadmill resulted in an intuitive change in activation levels of the anti-gravity muscles [17] examined, but again, the effect sizes seen were small in comparison to manipulating speed. Mindful of the inter-individual variations mentioned earlier, we suggest that for increasing muscle activity, relatively larger changes in indicated bodyweight and inclination can be administered.

4.3. Relatively fast downhill running has low activation of the posterior chain

The hamstring group of muscles appear to have relatively low activation during downhill running irrespective of the gradient, and only a slight increase as the positive gradient increases. In line with previous research, increasing running speed was associated with a strong increase in hamstring activation [5,10]. The gluteus maximus muscle appears to have a slightly stronger increase in activation for the increasing positive inclinations, although again, this effect is smaller than that seen for increasing running speed. Running downhill, however, resulted in a marked drop in gluteus maximus activation relatively independent of running speed. Similarly, the ankle plantarflexors showed a marked reduction in activation in the downhill conditions which appeared consistent across the three bodyweight conditions, and an intuitive increase in the positive inclinations. Again, running speed increases were much more strongly associated with increased activation here. These findings may be of interest during the rehabilitation of the hamstrings and plantarflexors (e.g. calf muscle injury, Achilles tendonopathy) where athletes may be able to more rapidly resume higher speed downhill running than level running while still appropriately loading these structures.

4.4. Cadence effects were more consistent than peak sEMG

Less between-subject and between-condition variability in cadence was seen in comparison to the EMG data. The strongest effect on increasing cadence was seen for increasing running speed (12 km/h: 83.71 mean strides per minute, 24 km/h: 104.74). In contrast to the EMG data, clearer effects on cadence were observed for bodyweight and inclination albeit with similarly reduced magnitude to the changes effected by increasing running speed. Higher cadence was routinely observed with increased bodyweight (60 %: 88.23, 100 %: 93.88) across all conditions. The lowest cadence was observed during level running (at the slowest speeds and lowest bodyweight) with higher cadence observed with both increasing positive and negative inclination however the effect of increasing positive (uphill) cadence was stronger (0%: 84.31, +15 %: 87.21) than downhill (-15 %: 84.76). Note however that downhill running only was only performed at 12 and 15 km/h.

4.5. Limitations

A clear gap in the data presented here are kinematics and kinetics of the lower limbs. The cadence data was not integrated with the EMG data, and no other measures of kinematics or kinetics are presented. Previous research has shown alterations in flight and contact time for different bodyweight conditions [18]. Muscle activation levels represent one aspect of load during activity, but these are significantly modified by the joint ranges through which they act as well as their velocity. Previous research [19,20] comparing flat and incline running has shown that running fast up an incline was associated with different ranges of motion at the hip, knee, and ankle, and a reduction in stance (but not swing) phase peak EMG activation of the hamstrings. Subsequent research documenting these data, and ideally calculating muscle and tendon forces would clarify the actual work done by these elements and better inform loading interventions. Only two speeds were analysed for the downhill conditions and given the likely non-linear association between muscle activation levels and running speed future research should examine extra conditions, ideally at higher speeds. Additionally, these data are for healthy recreationally active male runners and likely do not extrapolate to other populations.

5. Conclusions

Manipulating running speed, from 12 km/h up to 24 km/h results in much greater changes in EMG (both peak and iEMG) than does manipulating indicated bodyweight (60%–100%), or inclination (from downhill to uphill, -15 % to +15 %). An apparent transition from level to downhill is evident with a different pattern of lower, but more tightly clustered, activity across the downhill conditions for all muscles. Broadly, increasing indicated bodyweight is associated with increased activity, but the effect is relatively smaller, at least until the higher running speeds are considered. Understanding the relationships between muscle activation and these 3 treadmill loading parameters allow clinicians to effect similar loading changes via different means (i.e. speed, inclination, or relative bodyweight) as clinically indicated. Clinicians wishing to progress muscle activation levels during rehabilitation should be mindful that changes in running speed elicit relatively larger changes than seen when manipulating percentage bodyweight and inclination. Cadence changes were less variable than sEMG.

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Authors' statement

Each of the authors has read and concurs with the content in the final manuscript. The material within has not been and will not be submitted for publication elsewhere except as an abstract.

Declaration of Competing Interest

The authors report no declarations of interest.

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Appendix A. Supplementary data

Supplementary material related to this article can be found, in the online version, at doi:<https://doi.org/10.1016/j.gaitpost.2020.09.029>.

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