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Title: **Effects of wearable resistance load placement on neuromuscular activity and stride kinematics: A Preliminary Study**

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Abstract

Objectives: To investigate the acute neuromuscular and stride characteristic responses to different wearable resistance (WR) loads and placements on the calf muscles during high-speed running.

Methods: 10 well-trained subjects completed a workout of 10 sets of three 10-s runs at 18km.h^{-1} (20-s of rest between runs and 1-min between sets). Five conditions were tested: 1) unloaded control, 2) bilateral 0.75 vs. 1.5% body mass (BM) loading on the distal posterior calf, 3) bilateral proximal vs. distal loading of 1.5% BM positioned posteriorly, 4) bilateral anterior vs. posterior loading of 1.5% BM positioned distally, 5) unilateral loading of 1.5% BM on the distal, posterior calf. Data were collected using Electromyography (EMG) and back-mounted GPS-embedded accelerometers. Magnitude of differences of within-athlete, between-muscle comparisons were calculated using effect sizes (ES) ± 90% confidence limits (CL).

Results: No substantial differences in accelerometry data were observed between any of the loaded conditions and the control.

EMG activity was lower for proximal loading compared to control for the *Gluteus maximus* (ES±90%CL; -0.72±0.41), *Vastus lateralis* (-0.89±0.47) and *Vastus Medialis* (VM) (-0.97±0.46). Anterior loading induced substantially lower EMG activity for the *Semitendinosus* (-0.70±0.48) and VM (-0.64±0.39) muscles compared with control. EMG activity of the VM (-0.73±0.46) muscle was also substantially lower for posterior loading compared to control.

Unilateral loading induced no substantial differences in EMG activity between the loaded and unloaded legs.

Conclusions: This preliminary study has provided a rationale to perform further investigations into the effects of WR lower-limb loading on stride characteristics and EMG activity from a chronic standpoint using a larger population.

Keywords: Stride Characteristics, Neuromuscular Activity, Electromyography, Accelerometry, Muscle Activation

Introduction

A key for athletic performance is the successful transference of strength and power adaptations from gym-based exercises to sport-specific movements.[1] A practitioner's main goal when developing a resistance training program is to develop an athlete's strength and power outputs.[2, 3] However, those improvements might not always transfer to on-field performance with factors such as training methodology, volume, duration, and intensity influencing the transfer effect of training induced strength and power adaptations.[2, 3] To alleviate those inherent limitations, wearable resistance (WR) training may be a practical solution with a better ecological fit than traditional gym-based resistance training. WR training allows athletes to perform loaded sport-specific movements which are hypothesized to provide greater transference to sport-specific movement performance compared with traditional gym-based exercises.[1]

WR has been used extensively within athletics and sprint training.[1, 4] Research suggests WR can elicit increases in horizontal force production and improve sprint performance.[1, 4] Additionally, investigations have explored the potential performance benefits of using WR for team sports.[5-7] One study investigating the use of calf-loaded WR during warm-ups for elite footballers reported improvements in maximal horizontal (e.g., 10- and 20-m sprints) but not vertical (e.g., counter movement jump) performance.[5] In rugby players, the use of calf loaded WR (1% body mass, BM) resulted in a better maintenance of acceleration and sprint performance during a six-week training period compared to unloaded players.[6] Furthermore, WR training increased acute training load for high-school American footballers loaded with 1% BM on the calves.[7]

In addition to WR training effects on locomotor performance, there is growing knowledge regarding movement kinematics and muscle coordination involved with WR loading.[1, 8] During loaded sprinting, stride length and frequency decreased, while contact time and ground reaction forces increased.[1] Previous research showed that the positioning of WR impacts stride kinematics.[8] For example, greater kinematic changes were observed for calf-loaded WR conditions compared to thigh-loaded WR.[1] However, it is yet to be investigated how the effect of WR varies based on lower-limb load location (e.g., proximal *vs*. distal placement, anterior *vs*. posterior, bilateral/unilateral). If stride mechanics were to be altered, and/or specific muscle recruitment to be modified in relation to various WR placements (potentially affecting muscle coordination), this may have important implications for the integration of WR in training practices (e.g., rehabilitation).

Therefore, the aims of this preliminary study were to investigate changes in muscle activation amplitude and stride characteristics induced by the effects of WR calf loading of different loads (0.75%BM vs 1.5%BM), different load placements (anterior vs posterior and proximal vs distal), unilateral loading and loaded *vs*. unloaded conditions during high-speed running efforts.

Methods

Subjects

Ten well-trained male subjects (30.9±6.0yrs, 178.6±5.4cm, 75.8±5.8kg), who regularly partake in running and resistance-based training (8-hours per week) completed this study. Table 1 shows the anthropometric data of the participants. Subjects were free from injury and illness for at least 4-weeks prior to the start and gave consent for data obtained to be used in this study. Each participant gave consent for their data to be used and data collection was part of the club's regular monitoring procedures which conformed to the Declaration of Helsinki.[9]

Insert Table 1 here.

Intervention

Subjects completed a workout of 10 sets of three 10-s runs at 18 km.h^{-1} with 20-s rest between runs, and 1-min between sets. Each run was performed on a motorized treadmill (Skillrun Unity-7000, Technogym, Italy). Before each set, WR (Lila™ Exogen™, Malaysia) was placed on subject's lower-limbs. Five different WR conditions were tested: 1) control without load, 2) 0.75 vs. 1.5% BM loading positioned on the distal posterior calf, 3) proximal vs. distal loading of 1.5% BM positioned posteriorly 4) anterior vs. posterior loading of 1.5% BM positioned distally, 5) unilateral loading of 1.5% BM positioned on the distal, posterior calf (conditions 2-4 were bilaterally loaded). Figure 1 shows the experimental WR loading patterns. Data were collected using surface Electromyography (EMG) and an embedded accelerometer within an upper-back mounted GPS unit.

Measurements

Effect of load: 0.75% or 1.5% BM loading was placed on the posterior, distal portion of the lower-limbs, aligned with the *gastrocnemius* aponeurosis line of action (fig. 1A and B).

Anterior/Posterior: To assess the effect of anterior/posterior loading, a 1.5% BM load was placed distally either on the front (anterior) or rear (posterior) of the lower-limbs, aligned with the *tibialis anterior* insertion and the *gastrocnemius* aponeurosis line of action respectively (fig. 1B and D).

Proximal/Distal: To test the effect of proximal/distal loading, a 1.5% BM load was placed posteriorly either on the upper calf (proximal) or lower calf (distal) between the origins of the medial and lateral heads of the *gastrocnemius* and in alignment with the *gastrocnemius* aponeurosis respectively (fig. 1B and C).

Unilateral condition: The unilateral condition involved load placement on one leg with 1.5% BM placed on the posterior, distal portion of the lower-limb, aligned with the Gastrocnemius aponeurosis, while the other leg was unloaded (fig.1E). Participants completed the unilateral trial with both legs and this data was pooled to give an average unilateral measure.

Control condition: Running without additional load.

Insert Figure 1 here.

Data Collection Procedures

Electromyography. A BTS FREEEMG 300 wireless surface EMG system (BTS® Quincy, USA) was used with sensors placed on the *gluteus maximus* (Gmax), *bicep femoris* (BF), *semitendinosus* (SM), *rectus femoris* (RF), *vastus lateralis* (VL) and *vastus medialis* (VM) muscles for each leg. The skin was shaved, gently abraded and cleaned with alcohol to minimize inter-electrode impedance. The bipolar, silver/silver chloride, surface disc electrodes (Blue Sensor N-00-S/25; Ambu, Baltorpbakken, Denmark) were placed with a centre-to-centre distance of 2.5 cm, longitudinally with respect to the underlying muscle fiber arrangement and located according to the Surface EMG for the Non-Invasive Assessment of Muscle's (SENIAM) recommendations.

The sampling frequency was 1 000 Hz. The EMG data processing technique started with filtering the EMG signal (High pass, 15 Hz, third order Butterworth filter). Secondly, the calculation of the muscle activity (mean root mean square, RMS) during each of the trials was performed. The first and last few steps were excluded from the recording window to keep only the stable phase of the run. The treadmill was continually moving with participants instructed to 'jump' onto the side between trials. This minimized any acceleration required during each repetition in an aim to increase the stability of the runs. The mean RMS for each trial was calculated. The three trials for each condition were averaged (CV: 6.1-8.8%). The control condition mean RMS was used to normalize all other conditions. EMG data is displayed as a % of the normalized condition.

Stride characteristics. Embedded accelerometers (952 Hz) in GPS units (StatSports®, Northern Ireland) were used in indoor mode to calculate bilateral stride kinematics. Accelerometry raw data was further analyzed using ADI software (Athletic Data Innovations, Sydney, Australia) to derive floor contact time (CT [seconds]), peak force (Newtons), stride frequency (steps/second), and vertical stiffness (kvert [$KN.m^{-1}$]).(10) ADI-derived metrics were shown to be reliable with small to moderate error during high-speed running (standardized typical error:0.52-0.67).(10)

Data analysis. Muscle activity and accelerometry data for bilaterally loaded conditions were calculated using the average of three repetitions per condition from both legs and normalized in relation to the control condition (%). Muscle activity for the unilateral condition used the pooled average from the loaded legs (average of loaded left and right legs) to compare with the unloaded legs (average of unloaded left and right legs) to observe possible changes in muscle activation.

Statistics:

Data are presented as mean ± standard deviation (SD) and as effect size ± 90% confidence limits (CL). Data was first checked for normality (Shapiro-Wilks test). Log-transformation was used when required to transform skewed data to approximately conform to normality. Data was back transformed after analysis to return them back to original units. Within-athlete between-muscle comparisons were made using effect sizes based on Cohen's d principle to determine the magnitude of change between conditions (0.75 vs 1.5% BM, Anterior vs Posterior, Proximal vs Distal, Unilateral loading, Control compared with all conditions) using Hopkins' scale: 0.2 (small), 0.6 (moderate), 1.2 (large), 2.0 (very large).(11) When the CL of the ES didn't overlap the SWC (0.2), the change was considered substantial and of the observed magnitude; if the CL overlapped the SWC, the change was unclear.[12]

Results

Differences between conditions for accelerometry date are presented in Table 2 and differences between conditions for EMG data are presented in Table 3. Intra-subject between condition stride frequency data is displayed in Figure 2 showing large individual responses to the load and placement. Figure 3 shows the synchronization of left and right EMG data with vertical acceleration data for different loading patterns.

Insert Table 2 here.

Insert Table 3 here.

Effect of Load

No substantial differences were observed between 0.75% BM loading, 1.5% BM loading and the control for all stride characteristic metrics (all ES rated as unclear). Likewise, no substantial differences in stride characteristics were observed between 0.75% and 1.5% BM loading.

EMG activity for the VL and VM muscles were moderately lower for 0.75% BM loaded conditions compared with the control (ES:0.63-0.92). Additionally, 1.5% BM loading induced moderate decreases in EMG activity of the VM compared to the control (ES±90%CL; 0.70±0.44).

Anterior vs. Posterior Load Placement

Accelerometry data showed no substantial differences for all stride characteristics of the anterior and posterior conditions compared to the control (all ES rated as unclear).

Anterior loading induced moderately lower EMG activity for the ST and VM muscles compared with the control (ST:0.70±0.48, VM:0.64±0.39). EMG activity of the VM muscle was also moderately lower for posterior loading compared to the control (0.73±0.46).

No substantial differences in stride frequency were observed between anterior and posterior loading. Furthermore, no substantial differences in EMG activity were observed for any muscle between anterior and posterior loading (all ES rated as unclear).

Proximal vs. Distal Load Placement

Proximal and distal loading conditions showed no substantial differences in stride characteristics compared with the control. Moreover, accelerometry data showed no substantial differences in stride characteristics were observed between proximal and distal loading (all ES rated as unclear).

The EMG activity of the Gmax, VL and VM was moderately lower for proximal loading compared to the control while the EMG activity of the VM was moderately lower for distal loading compared to the control (ES:0.65-0.97).

Unilateral Loading

For unilaterally loaded conditions no substantial differences in EMG activity were observed between the loaded and unloaded leg (all ES rated as unclear).

Insert Figure 2 here.

Insert Figure 3 here.

Discussion

The aims of this preliminary study were to investigate the effects of different WR loads and loading placements during high-speed running efforts on stride characteristic variables and EMG responses. The main findings were as follows: 1) Bilateral WR loading induced no substantial changes in stride characteristics or force metrics for all loads and placements, 2) Proximal loading patterns moderately decreased Gmax, VL and VM EMG activity while distal loading patterns induced moderate decreases in VM EMG activity, 3) Anterior and posterior WR load placement induced decreases in ST and VM EMG activity.

Overall loading effects

Accelerometry data showed no substantial changes in stride characteristics (CT, stride frequency) and force metrics (peak force, Kvert) of the loaded conditions compared with the control. This study does not support previous findings regarding the effects of WR on stride characteristics and force metrics.[1] Previous research highlighted that WR induced increases in stride frequency can occur in parallel to decreases in stride length,[13] which may result in decreased lower-limb muscle activity.[14] However, as the observed changes in stride frequency were unsubstantial, changes in muscle activity may have been as a result of participants changing their joint kinematics, which may change the amount of muscle under the electrode thus, potentially changing EMG amplitudes and recorded muscle activity.[1] Regarding stride frequency, there were large differences in individual responses to the load and placements which are difficult to explain without EMG and accelerometry synchronization but would be worth investigating in future studies. Previous studies have shown WR to increase the metabolic load of training.[15] This, in addition to its ability to decrease neuromuscular load, may highlight the ability of WR to be a useful tool for coaches to utilize for training purposes while minimizing injury risks. However, further research is necessary to quantify the effects of calf loaded WR on stride kinematics.

Effect of the load

When examining the effect of load, when compared to the control there were no substantial effects of 0.75% or 1.5% BM WR loading on stride characteristics or force metrics. Additionally, no differences were observed from the accelerometry data between the two loaded conditions. However, we found that EMG activity of the quadricep muscles (VL and VM) were substantially lower for the 0.75% BM loaded condition compared to the control. Furthermore, the VM muscle showed decreased EMG activity for the 1.5% BM condition compared to the control. The VL and VM are hip flexors responsible for force production and the stabilization of the knee during running[16, 17] and exhibit their highest workload during the foot strike and early stance phases of running.[16] These finding suggests that 0.75% and 1.5% BM loading induced decreases in neuromuscular workload for quadricep muscles responsible for force production during running without effecting running mechanics. Therefore, WR could potentially be a useful tool for rehabilitation protocols to reduce neuromuscular load and potentially minimize injury risk. However, further research is required to investigate these findings using a larger testing cohort.

The finding of no substantial differences in accelerometry data between the two loaded conditions supports previous findings whereby there were no meaningful changes in stride frequency between 3% and 5% BM lower-body loading.[1] Contrary to the previous study in which the load was distributed on the thighs and calves, the load in this study was focused solely on the calves. Calf loading can induce a greater rotational inertia than thigh loaded WR due to its increased distance from the rotational centre (hip joint)[1] thus, increases in calf loading could potentially result in greater effects to stride frequency than thigh loading. However, this needs to be investigated using a larger scale study. This information may be useful for coaches wanting to utilize WR loading patterns that may maximize performance adaptations elicited by this training modality. Interestingly, EMG data showed no changes in muscle activity between the 0.75% and 1.5% BM loaded conditions. This was potentially due to minimal increases in motor unit recruitment from the small loads relative to BM. Alternatively, in accordance with previous studies, the lack of EMG changes may have been due to a reduced EMG sensitivity to small differences in loads.[18-20]

Effect of load placement

Regarding load placement, we found that anterior, posterior, proximal and distal load placements did not have any clear effects on stride characteristics. Previous research investigated the acute kinetic changes of shank vs thigh loading,[1] but to our knowledge, this is the first study to investigate varying calf loading patterns. There were no meaningful differences found between posterior and anterior 3% BM thigh-loaded WR during sprint accelerations for kinematic measurements.[1] However, we found that proximal loading caused decreases in Gmax, VL and VM EMG activity compared with the control, and distal, anterior and posterior loading also caused decreases in VM EMG activity, whereas no changes in stride frequency were observed between these conditions. The roles of the VL and VM muscles during running were previously stated and due to their importance for force production, proximally placed lower-limb WR loading may be a useful tool to reduce the neuromuscular workload of these muscles and reduce injury risks.[16, 17] Moreover, due to their role as hip flexors involved in vertical movement patterns such as raising the leg during running, the decreases in VL and VM EMG activity may suggest an association with decreases in stride length.[17] However, further investigations are required to research these points in the context of WR training.

As previously stated, decreases in Gmax EMG activity was induced by proximal loading. The major functions of the *gluteus maximus* muscles during running are to provide trunk stability during the stance phase, decelerate the swing leg and assist with leg extension.[21] The Gmax is most active during high-speed running.[21] Decreases in Gmax EMG activity may be useful for reducing neuromuscular load during high-speed running thus, proving to be a useful tool for rehabilitation training by reducing potential injury risks. However, as stated previously, further research is required to investigate these findings.

Unilateral loading

Unilateral WR loading showed no substantial differences in EMG activity between the loaded and unloaded leg. As previously discussed, it is possible the lack of substantial differences observed between the loaded and unloaded leg may have been due to a reduced EMG sensitivity to these small loads.[18] Further research is required to assess the effects of WR unilateral loading on stride characteristic and EMG activity using greater loads.

While the study findings may not have been conclusive in identifying the stride characteristics and EMG activity of using lower-limb WR, the results from previous lower-limb loaded WR studies imply that using WR may allow coaches to induce a training overload specific to sport specific movement mechanics.[5-7] Furthermore, lower-limb WR can possibly be used to increase acute training workloads, but further research is required with larger sample sizes to further understand the locomotor and neuromuscular effects of WR training.[6, 7] Future research may also aim to increase the WR load or increase exposure times to the load, to investigate potential performance benefits of WR training. From a chronic standpoint, it is not known what performance adaptations a prolonged period of WR training could potentially provide, thus, further research will be required to investigate this.

Limitations

The main limitation of this study was the lack of mechanical and EMG data synchronization which would allow us to observe EMG differences of different stride phases. While using a within activity normalization approach for a high-speed, highly dynamic activity seems preferable, another approach would have been to perform maximal voluntary contractions (MVC) of each lower-limb muscle before testing and using this to normalize EMG signals across all conditions. This would have ensured accurate comparisons of intra- and inter-muscle activity. Furthermore, it was not possible to measure unilateral accelerometry data to identify unilateral load placement effects on stride kinematics. Finally, the sample size of this preliminary study is small thus, lacks statistical power to draw general conclusions, but its value allows for the implementation of larger WR training studies with greater statistical power.

Conclusions

In conclusion, this preliminary study suggests WR induces locomotor and neuromuscular change during high-speed running. These findings have provided the rationale to design further research studies using larger sample sizes to investigate what stride characteristic and neuromuscular performance adaptations can be provided by acute and chronic exposure to WR training.

Author Contributions

All listed authors contributed substantially to the concept, design analysis and interpretation of data and all authors approve the version to be published.

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Tables

Subject	Age (yrs)	Height (cm)	Body Mass (kg)
1	25.0	185	80.9
2	23.2	175	68.3
3	35.2	178	75.5
4	31.2	170	68.0
5	23.5	181	77.2
6	26.0	172	69.7
7	41.4	183	73.7
8	37.4	188	83.6
9	33.3	176	85.1
10	33.1	178	76.0
Mean ± SD	30.9 ± 6.0	178.6 ± 5.4	75.8 ± 5.8

Table 1. Individual subject anthropometric data: age (yrs), height (cm) and body mass (kg).

	Mean ± SD			Effect Size ± 90% CL		
	Control	**0.75%**	**1.50%**	**Control vs 0.75%**	**Control vs 1.5%**	**0.75 vs 1.5%**
Contact Time (ms)	0.17 ± 0.01	0.17 ± 0.01	0.17 ± 0.01	0.12 ± 0.28	0.11 ± 0.22	0.00 ± 0.20
Peak Force (N)	4 439 ± 299	4 448 ± 251	4 499 ± 232	0.04 ± 0.19	0.22 ± 0.24	0.18 ± 0.20
Freq (step/s)	2.99 ± 0.10	3.01 ± 0.15	3.07 ± 0.10	0.08 ± 0.48	0.53 ± 0.44	0.45 ± 0.43
kVert (KN.m^{-1})	121 ± 14.9	120 ± 15.4	121 ± 15.9	-0.09 ± 0.33	0.00 ± 0.28	0.09 ± 0.17
	Control	**Ant**	**Post**	**Control vs Ant**	**Control vs Post**	**Ant vs Post**
Contact Time (ms)	0.17 ± 0.01	0.17 ± 0.01	0.17 ± 0.01	0.27 ± 0.75	0.12 ± 0.22	-0.15 ± 0.95
Peak Force (N)	4 439 ± 299	4 492 ± 231	4 499 ± 232	0.38 ± 0.50	0.23 ± 0.24	-0.09 ± 0.47
Freq (step/s)	2.99 ± 0.10	3.06 ± 0.12	3.07 ± 0.10	0.67 ± 0.50	0.59 ± 0.49	0.06 ± 0.78
kVert (KN.m^{-1})	121 ± 14.9	120 ± 16.3	121 ± 15.9	-0.50 ± 0.57	0.00 ± 0.27	0.07 ± 0.81
	Control	**Prox**	**Dist**	**Control vs Prox**	**Control vs Dist**	**Prox vs Dist**
Contact Time (ms)	0.17 ± 0.01	0.17 ± 0.01	0.17 ± 0.01	0.17 ± 0.20	0.11 ± 0.22	-0.06 ± 0.18
Peak Force (N)	4 439 ± 299	4 531 ± 238	4 499 ± 232	0.34 ± 0.45	0.23 ± 0.24	-0.11 ± 0.45
Freq (step/s)	2.99 ± 0.10	3.08 ± 0.17	3.07 ± 0.10	0.60 ± 0.64	0.52 ± 0.43	-0.08 ± 0.58
kVert (KN.m^{-1})	121 ± 14.9	122 ± 17.7	121 ± 15.9	0.02 ±0.24	0.00 ± 0.27	-0.01 ± 0.20

Table 2. Accelerometry data: Mean ± Standard Deviation; Effect Size ± 90% Confidence Limits. Bilaterally loaded conditions – 0.75 vs 1.5% BM loading, Anterior vs Posterior loading, Proximal vs Distal loading.

	% Normalized condition ± SD		Effect Size ± 90%CL		
	0.75%	1.50%	0.75% vs 1.5%	Control vs 0.75%	Control vs 1.5%
BF	93 ± 36	93 ± 36	-0.04 ± 0.17	-0.34 ± 0.25	-0.38 ± 0.27
Gmax	71 ± 50	71 ± 50	0.01 ± 0.31	-0.50 ± 0.46	-0.50 ± 0.39
RF	83 ± 70	75 ± 67	-0.09 ± 0.24	-0.34 ± 0.29	-0.42 ± 0.33
ST	87 ± 31	80 ± 33	-0.16 ± 0.53	-0.47 ± 0.50	-0.63 ± 0.55
VL	75 ± 42	81 ± 39	0.11 ± 0.19	-0.63 ± 0.34*	-0.52 ± 0.41
VM	71 ± 50	77 ± 46	0.22 ± 0.18	-0.92 ± 0.40*	-0.70 ± 0.44*
	Ant	Post	Ant vs Post	Control vs Ant	Control vs Post
BF	93 ± 36	93 ± 36	0.11 ± 0.15	-0.26 ± 0.30	-0.37 ± 0.26
Gmax	71 ± 50	71 ± 50	-0.13 ± 0.17	-0.39 ± 0.46	-0.52 ± 0.40
RF	92 ± 63	75 ± 67	-0.21 ± 0.26	-0.21 ± 0.30	-0.42 ± 0.33
ST	80 ± 25	80 ± 33	0.05 ± 0.28	-0.70 ± 0.48*	-0.65 ± 0.57
VL	81 ± 39	81 ± 39	-0.05 ± 0.16	-0.48 ± 0.35	-0.53 ± 0.42
VM	77 ± 46	77 ± 46	-0.08 ± 0.27	-0.64 ± 0.39*	-0.73 ± 0.46*
	Prox	Dist	Prox vs Dist	Control vs Prox	Control vs Dist
BF	87 ± 39	93 ± 36	0.10 ± 0.25	-0.47 ± 0.31	-0.38 ± 0.27
Gmax	64 ± 44	71 ± 50	0.22 ± 0.36	-0.72 ± 0.41*	-0.50 ± 0.39
RF	75 ± 67	75 ± 67	-0.07 ± 0.22	-0.37 ± 0.30	-0.45 ± 0.35
ST	80 ± 33	80 ± 33	0.08 ± 0.39	-0.72 ± 0.59	-0.64 ± 0.56
VL	69 ± 45	81 ± 39	0.43 ± 0.29	-0.89 ± 0.47*	-0.46 ± 0.36
VM	65 ± 55	77 ± 46	0.32 ± 0.27	-0.97 ± 0.46*	-0.65 ± 0.41*

Table 3. EMG data: % normalized condition ± Standard Deviation; Effect Size ± 90% Confidence Limits. Bilaterally loaded conditions – 0.75 vs 1.5% BM loading, Anterior vs Posterior loading, Proximal vs Distal loading. * Represents a substantial difference compared with the other condition. *Gluteus maximus* (Gmax), *bicep femoris* (BF), *semitendinosus* (SM), *rectus femoris* (RF), *vastus lateralis* (VL) and *vastus medialis* (VM).

Figures

Figure 1. The experimental conditions – WR lower limb loading patterns. Pictures from left to right: A) 0.75% BM distal, posterior loading; B) 1.5% BM distal, posterior loading; C) 1.5% BM proximal, posterior loading; D) 1.5% BM distal, anterior loading; E) 1.5% BM distal, posterior, unilateral loading.

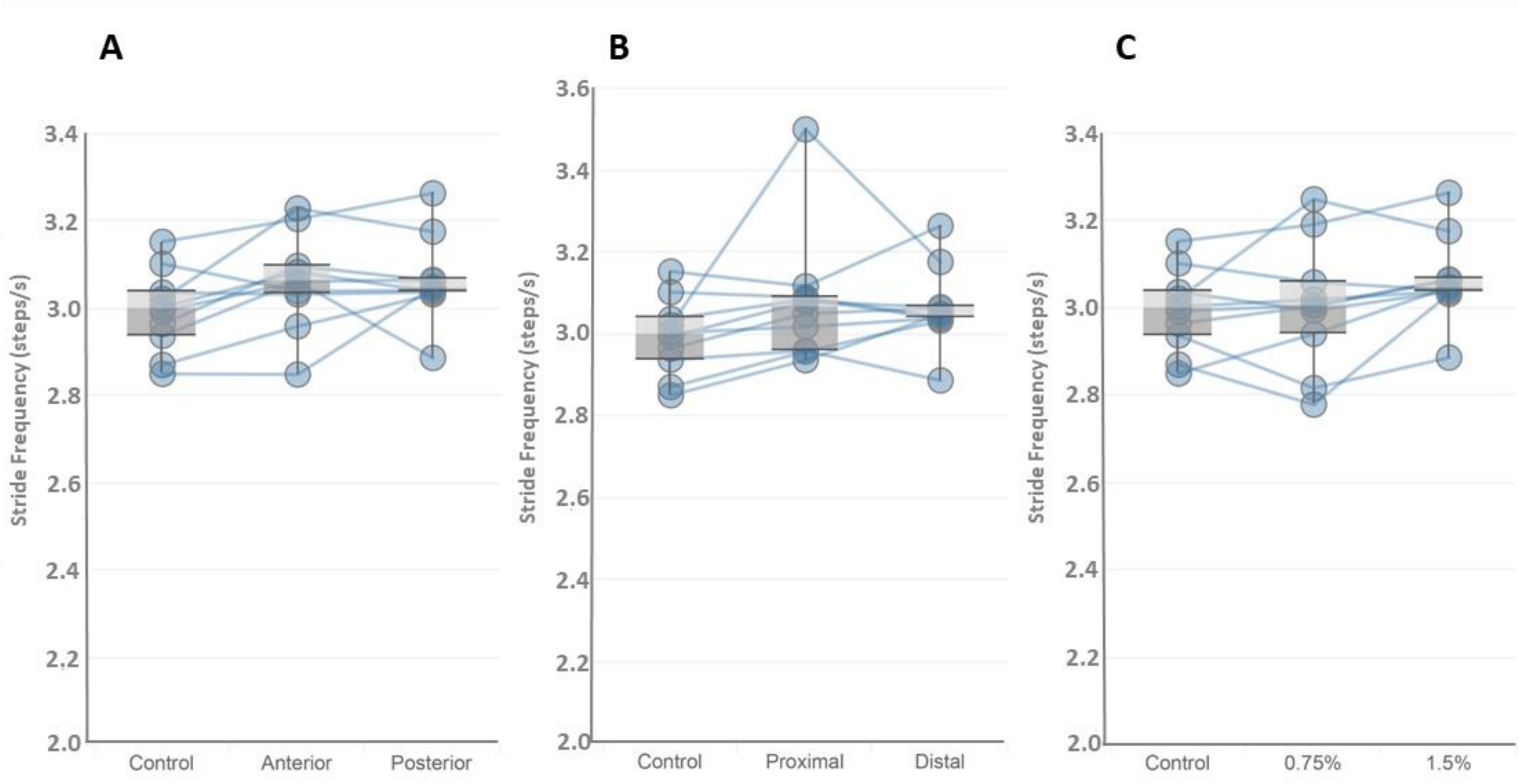

Figure 2. Intra-subject stride frequency (steps/sec) values for each WR load and loading pattern with group statistics shown within box plots. Each subject is represented by a line on each graph. A) Anterior and posterior loading compared to control, B) Proximal and distal loading compared to control, C) 0.75% and 1.5% BM loads compared to control.

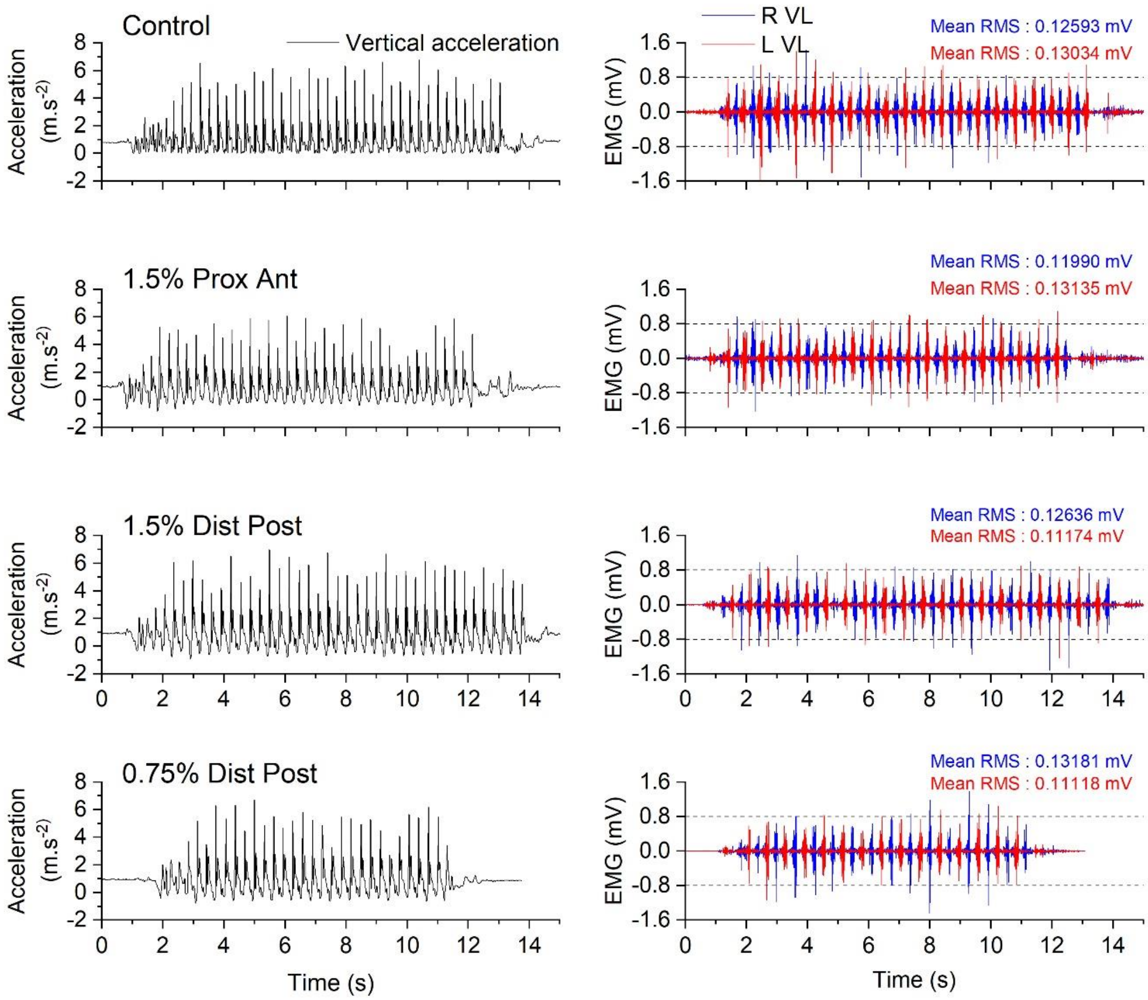

Figure 3. Typical raw data for vertical acceleration and EMG activity for four different conditions. EMG activity is expressed for the right (R) and left (L) vastus lateralis (VL). The mean root mean square (RMS) is indicated for each muscle and each trial presented. The four presented conditions are: control condition; 1.5% BM proximal, anterior loading; 1.5% BM distal, posterior loading; 0.75% BM distal, posterior loading.