

Effects of Resistance and Speed on Electromyographic Activity of Thigh and Gluteal Muscles in Elite Athletes Throughout Resisted Sprint Running

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ABSTRACT

MACCHI, R., A. HEGYI, C. GIROUX, G. RABITA, A. NORDEZ, E. HOLLVILLE, A. COUTURIER, C. FORNASIER-SANTOS, J. SLAWINSKI, J.-B. MORIN, L. LACOURPAILLE, and G. GUILHEM. Effects of Resistance and Speed on Electromyographic Activity of Thigh and Gluteal Muscles in Elite Athletes Throughout Resisted Sprint Running. *Med. Sci. Sports Exerc.*, Vol. 57, No. 7, pp. 1530–1545, 2025. **Purpose:** The application of horizontal resistance is key in sprint training to modulate velocity and promote adaptations. However, the interaction between velocity and resistance on muscle excitation remains insufficiently studied. This study assessed the electromyographic (EMG) activity of thigh and gluteal muscles in response to varying velocity and resistance during sprinting. **Methods:** Thirty-seven elite athletes (27 females and 10 males) performed two trials under three conditions: a 40-m maximal sprint without resistance and two resisted sprints applied by a robotic device with resistive forces equivalent to 25% and 75% of body mass in a sled condition. EMG activity was recorded from eight muscles in the quadriceps, hamstrings, and gluteal muscle groups for both lower limbs. Generalized mixed models were used to analyze average EMG changes with increasing velocity as a function of resistance. Statistical parametric mapping was used to assess the changes within the stance and swing phases as a function of sprint phase and resistance. **Results:** Hamstring EMG activity increased with increasing velocity (+17.9% in stance phase without resistance) and decreased as resistance increased, with opposite effects observed in quadriceps muscles. Higher gluteal EMG activity was found during the late swing phase with increasing velocity and decreasing resistance. The biceps femoris long head exhibited the largest EMG increase with velocity (+23% in the stance phase) among hamstrings, whereas the semitendinosus showed the highest increase with reduced resistance (+27% in the stance phase). **Conclusions:** External resistance and running speed influence thigh muscle activity differently, even within the same muscle group, likely due to distinct muscle architecture and function. These insights should be considered when designing sprint training programs to target specific muscle groups, avoid muscle overloading, and account for interindividual differences to optimize performance and reduce injury risk. **Key Words:** BICEPS FEMORIS, EMG, LOCOMOTION, SEMIMEMBRANOSUS, SEMITENDINOSUS, SPORT PERFORMANCE

The ability to sprint is a key factor for success in many sports. Acceleration during sprinting is generally divided into three successive phases derived from the kinetics of the center of mass: early acceleration, late acceleration, and maximum speed (1–3). Each sprint phase necessitates

the recruitment of specific muscles, prompting several studies to investigate the effects of velocity on muscle excitation through surface electromyographic analysis (e.g., [4,5]).

Previous studies have shown that hamstring electromyographic (EMG) activity is higher, whereas quadriceps EMG activity is lower at high running speeds compared with low speeds, with distinct differences in the stance and swing phases (4–9). In these studies, running tasks were conducted at steady speeds. During a maximal sprint acceleration on a treadmill, an increase in hamstring EMG activity was reported during the first half of the stance phase and the end of the swing phase (10). On the contrary, quadriceps EMG activity decreased only in the first half of the stance phase, whereas no change was observed for the gluteus maximus (GMax) (10). To complement these studies, further research is required to comprehensively examine (i) the impact of sprint acceleration on EMG activity

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Submitted for publication November 2024.

Accepted for publication February 2025.

0195-9131/25/5707-1530/0

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DOI: 10.1249/MSS.0000000000003689

by differentiating various phases of a single maximal sprint and the running cycle (i.e., stance and swing), and (ii) the excitation patterns of multiple muscles within the same muscle group.

Resisted sprinting is a widely used training method to improve propulsion in the acceleration phase by mimicking high horizontal force applied to the ground during the first steps of sprint acceleration (11,12). Previous studies found that mean EMG activity decreased in the hamstrings and increased in the quadriceps in response to increasing resistance (13–16). These global changes may conceal distinct variations depending on the stride cycle or sprint phase, requiring further research to enhance our understanding of the effects of resistance on EMG activity. In addition, no study has specifically examined the between-individual consistency in EMG responses to changes in speed and resistance. Given the importance of individualized training for elite athletes, quantifying this variability could help determine whether assessments should be customized for each athlete. Finally, little is known about the changes in EMG activity in the different heads of a given muscle group to running velocity and resistance. In this view, biceps femoris long head (BFLh) is the most frequently examined hamstring muscle, and some studies examined medial hamstrings as one functional unit, without distinguishing between semitendinosus (ST) and semimembranosus (SM). This appears as a reductionist approach considering that each hamstring muscle has distinct mechanical properties and architecture (17,18). For example, the fusiform architecture of ST is recognized to have a mechanical advantage when hamstrings shorten relatively fast, whereas SM, which is a pennated muscle, has a mechanical advantage at relatively slower shortening velocities compared with other hamstring muscles. This suggests that a redistribution of EMG activity within the hamstring muscle group could be needed to optimize the contribution of each muscle to the global joint torque and ultimately to sprint performance according to the instantaneous running speed and resistance.

The aims of this study were (i) to model mean changes in EMG activity in response to increasing resistance and normalized sprint speed in elite athletes, (ii) to conduct a detailed analysis of EMG activity throughout the running cycle, (iii) to assess between-individual consistency in EMG activity to evaluate the need for individualized field measurements, and (iv) to compare changes in EMG activity between hamstring muscle heads to investigate potential redistribution of EMG activity in response to variations in sprint speed and resistance.

We hypothesized that with increasing velocity and decreasing resistance, hamstring EMG activity would increase while quadriceps EMG activity would decrease, whereas these changes would vary depending on the phase of the sprint cycle. Furthermore, we expected a higher hamstring EMG activity during the late swing and early stance phases, when the hamstrings are highly activated. Considering the important role of hamstrings at high running speeds, we hypothesized that hamstring EMG activity would increase in a substantial proportion of individuals as speed increases and resistance decreases. Finally, we expected significant differences among

the hamstring muscles in their responses to changes in velocity and resistance, based on their architectural configurations.

METHODS

Participants

This study initially involved 49 elite athletes, defined as those competing at the international level according to McKay et al. (19). Some participants were excluded due to muscle pain during testing, poor EMG signal quality, or issues with synchronization between measurement tools, leading to the final inclusion of 37 athletes (27 females and 10 males; age: 27.0 ± 3.3 yr, height: 1.74 ± 0.10 m, body mass (BM): 70.9 ± 13.2 kg), which were considered for further analysis. Participants included elite track and field (10 females and 5 males), rugby seven (14 females and 5 males), and bobsleigh (3 females) athletes. Their maximal velocity over a 40-m sprint was 8.66 ± 0.55 m·s⁻¹ (min = 7.70, max = 9.68) in females and 9.96 ± 0.46 m·s⁻¹ (min = 9.19, max = 10.66) in males. Participants were injury free. They frequently performed maximal sprint accelerations as part of their usual weekly training program. The project was approved by the French Research Ethics Committee (No. 2021-A02523-38), and written informed consent was obtained from all athletes in accordance with the Helsinki Declaration.

Experimental Design

Each athlete was firstly equipped with surface EMG electrodes before performing an individual 20–45 min warm-up (mobility, athletic drills, and accelerations). Then they performed two trials of three sprint conditions on an official indoor track (tartan®), using a three-point start (i.e., with the contralateral hand of the front leg placed on the track). Athletes were instructed to refrain from consuming caffeine during the experimental protocol and run in their usual shoes. The front leg and rear leg were always the same (right or left lower limb according to preference). The conditions were performed in the following order: an unresisted condition consisting of a 40-m maximal running sprint (R0) and two resisted conditions consisting of 25- and 20-m sprints with resistances equal to 25% (R25) and 75% (R75) of BM, respectively. These resistances corresponded to the relative BM added in a sled device condition and did not represent horizontal resistance (see the Measurement Protocol section). The selected resistances were chosen to ensure sufficiently distinct conditions in terms of the induced velocity decrement and to correspond to proposed resistances in individualized resisted sprint training (20). Following the terminology proposed by Cahill et al. (21), R25 was designed to target the speed–strength portion (i.e., 20% of decrement in maximal velocity) of the force–velocity relationship, whereas R75 focused on the power production portion (i.e., 50% of decrement in maximal velocity). The sprint distances were adjusted to allow for an adequate number of steps within the targeted speed zone. For each athlete, the distance was systematically checked to ensure

the achievement of at least 95% of maximal running speed. Each sprint was separated by a recovery period of 4 min. All athletes were familiarized with the test procedures and performed one trial of each sprint condition during the warm-up.

Measurement Protocol

The resisted conditions were elicited by a portable robotic resistance device (1080 Sprint; 1080 Motion, Lidingö, Sweden). The motor was attached to a composite fiber cord that is wrapped around a spool, and attached to the athlete by a hip harness around the pelvis. The selected resistance levels on the robotic device corresponded to loads of 25% and 75% of BM as applied in sled conditions. A conversion coefficient of 0.35, provided by the manufacturer, was used for this calculation. The actual horizontal resistances induced by the device were therefore 8.75% BM and 26.25% BM. For example, if an athlete weighing 80 kg requires a sled resistance of 75% BM (i.e., a 60-kg load), the horizontal resistance programmed into the machine would be set to 21 kg (i.e., 0.35×60). The robotic device placed on the track at 5 m behind the starting position was set in isotonic mode in order to apply a constant resistance independently of acceleration, thus minimizing inertia.

Vertical, anteroposterior, and lateral components of the ground reaction forces were recorded along a 6.60-m-long force plate system sampling at 1000 Hz (KI 9067; Kistler, Winterthur, Switzerland). This system consisted of six individual force plates connected in series and covered with a tartan mat that was leveled with the stadium track. The second force plate was turned by 90° to allow the athletes' hands positioning during the start. The force plate system allowed to identify the sprint start event and to detect the touchdown and toe-off events. The force plates were synchronized with an optoelectronic system (Optojump Next; Microgate, Bolzano, Italy) positioned at floor level over a 40-m distance (comprising 80 parallel bars: 40 receivers and 40 transmitters) using the first touchdown after the sprint start as the synchronization point. The optoelectronic system was used to identify the touchdown and toe-off of all steps, except for the first steps, which were detected by force plates (this included the first three to five steps). As the toe-off event of the rear foot during the sprint start was not possible to detect using force signals, a GoPro camera (GoPro Inc, San Mateo, CA) was used to detect it using Kinovea® application version 0.9.5. The time between the toe-off of the rear foot and the first touchdown of the same foot on the force platform was calculated. This allowed us to identify the instant of the rear foot's toe-off on the force signal.

Surface EMG activity was recorded from eight muscles of each leg (double differential, 10-mm interelectrode distance, Trigno Wireless System; Delsys Inc, Boston, MA): GMax, gluteus medius (GMed), SM, ST, BFlh, biceps femoris short head (BFsh), vastus lateralis (VL), and rectus femoris (RF). Skin preparation and position of the bipolar electrodes followed SENIAM recommendations (22). To differentiate

the hamstring heads and to minimize crosstalk, the appropriate location of the electrodes was carefully determined using ultrasound scanning in B-mode (Aixplorer Mach 30 and Linear SL18-5; Supersonic Imagine, Aix-en-Provence, France) before electrode placement at rest in a prone position. The EMG signals were recorded continuously during the sprints at 2148 Hz and were synchronized by an electronic trigger delivered to a specific channel synchronously recorded with force signals from the force plates.

Data Processing

Velocity signal. For each sprint, the velocity signal was computed from the optoelectronic system, by dividing the distance between two steps by the stride time. Then the velocity over time was fitted with an exponential function using the least-square regression method (23).

$$V_H(t) = V_{Hmax} \cdot (1 - e^{-(t-d)/\tau})$$

where V_{Hmax} is the maximal velocity (in $m \cdot s^{-1}$), τ is the acceleration time constant (in $m \cdot s^{-1}$), and d is the time delay. To ensure the reliability of the optoelectronic system, we compared the inferred running velocity with the velocity obtained from a radar system (Stalker Pro II Sports Radar Gun, Plano, TX; 46.875 Hz) as a reference. The radar was positioned 5 m behind the starting position and 1 m above ground level. The analysis of unresisted sprints was divided into three distinct phases: early acceleration, late acceleration, and maximum speed. The τ values were employed to delineate the transition between early and late acceleration phases. A threshold of 5% of the maximum speed value was selected to separate the transition from late acceleration to the maximum speed phase. In the context of resisted sprint conditions, where the velocity pattern exhibits slight variations, two main phases were identified: acceleration and maximum speed.

Running cycle kinematics. The force signals were low-pass filtered (cutoff frequency, 200 Hz) using a third-order IIR Butterworth zero-phase filter. The sprint start was identified as an increase in vertical force for 15 consecutive milliseconds. The instants corresponding to successive touchdowns and toe-offs during force plate steps were identified as the inflection points where the normal component of the ground reaction force exceeded 20 N. A minimum separation of 150 ms between consecutive foot strikes was enforced to ensure distinct steps. For steps performed beyond the force plates, the touchdown and toe-off events were identified using the stance and flight times as recorded by the optoelectronic system.

EMG activity. The raw EMG signals (illustrated in Fig. 1) were band-pass filtered with cutoff frequencies of 10 and 450 Hz. Then, the signals were full-wave rectified. To obtain a linear envelope, each stance and swing phase was separately low-pass filtered using a fourth-order IIR Butterworth zero-phase filter. Based on the principles of the Shannon–Nyquist theorem, the cutoff frequency of the low-pass filter was meticulously adjusted according to the duration of each stance and swing phase (cutoff frequency: 19.7 ± 0.7 Hz). The cutoff

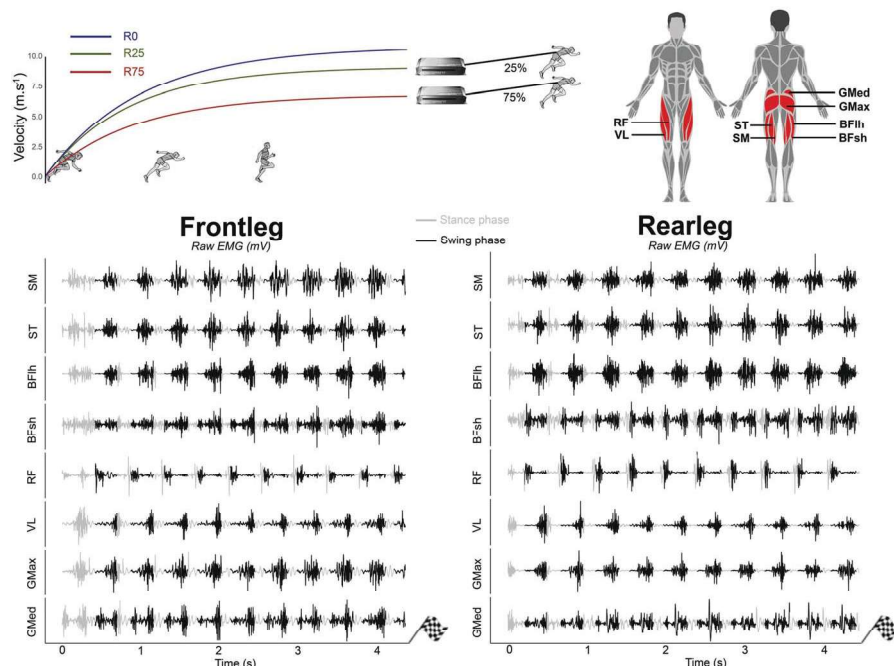


FIGURE 1—Illustration of fitted velocity signals (mono-exponential function) for the three resistance conditions and typical raw EMG signals from the lower limb muscles of both legs for an athlete. Although the sprints were not of the same duration between the different resistance conditions, the velocity signals of the exponential models have been extended in this visual example so that each curve corresponds to a duration of 4.5 s on the x-axis.

frequency was scaled by the ratio of the average contact time of a running cycle phase (i.e., stance or swing) to the duration of the signal under consideration. This adjustment ensured that the relative frequency characteristics of the filter remained consistent across different signal durations. The EMG patterns were averaged for each stance and swing phase according to the sprint phase. The maximal EMG activity obtained between the three phases was then used to normalize the EMG amplitude of all trials, regardless of the resistance condition, to allow for appropriate comparison between conditions. Finally, each stance and swing phase was time-normalized over 101 evenly separated points to allow interparticipant comparisons. The EMG signal for each running cycle was visually inspected before performing subsequent analysis.

Statistical Analysis

Statistics were performed using custom R scripts (R v 4.2.2, R Core Team, 2022; R Foundation for Statistical Computing, Vienna, Austria). The intertrial reproducibility of the EMG activity amplitude was assessed on the basis of the two sprint trials for each resisted condition and each muscle of each leg separately, taking into account the running cycle phase as well as the sprint phase. To achieve this, the root mean square error (RMSE) was computed for each individual separately and then averaged. The reproducibility error was compared with the interindividual variability, which was computed as the standard deviation across individuals.

EMG activity modeling as a function of normalized speed. For each muscle and each phase of the cycle (i.e., stance or swing), generalized mixed models were performed

using the glmmTMB package. These models included the EMG activity of both lower limbs. The most appropriate distribution family (e.g., Gaussian, lognormal, gamma, etc.) of the dependent variable (i.e., EMG activity) was selected for each model. The relationships between variables were analyzed to select the most appropriate fit for each model (e.g., linear, polynomial or exponential) using the Akaike information criterion (24). The normality of the residuals was carefully checked by Q-Q plots. Restricted maximum likelihood estimation was used in the model. The intercepts for the participants were chosen as random effect. The coefficients of the fixed effects were tested according to the Wald Z statistics.

Influence of stride cycle on EMG activity as a function of resistance and sprint phase. For both stance and swing phases, two separate statistical parametric mapping (SPM) analyses (25) were conducted for each muscle. The first analysis focused solely on the unresisted sprint condition (one-way repeated-measures ANOVA), considering the sprint phase as a repeated factor with three phases (i.e., early acceleration, late acceleration, and maximum speed). The second SPM analysis aimed to analyze the effect of resistance as a function of the sprint phase (two-way repeated-measures ANOVA), incorporating two sprint phases (i.e., acceleration and maximum speed) and three resisted conditions (i.e., R0, R25, and R75) as repeated factors. Holm's *post-hoc* test was used to perform pairwise comparisons. An additional SPM analysis was conducted, with sprint phase and leg (i.e., front leg or rear leg) included as repeated factors. Note that, given the absence of significant differences in EMG activity between the front and rear leg, apart from minor changes during the early acceleration phase, we focused the SPM analyses exclusively on the rear leg.

Between-individual consistency of the effects of velocity and resistance on EMG activity. For the velocity effect, we computed the difference between normalized EMG activity achieved at $V_{\max}$ minus the normalized EMG activity at minimal velocity ($V_{\min}$) for each resistance condition. To determine the influence of resisted condition, the difference in normalized EMG activity achieved at $V_{\max}$ was also calculated for each pair of condition comparison (i.e., R25 vs R0, R75 vs R0, R75 vs R25). For these two analyses, to address potential biases stemming from noise or other factors inherent in a single EMG value, we used a nonparametric approach (locally weighted scatterplot smoothing) (26) to obtain new average EMG activities for each individual. This method captures trends rather than relying on individual data points, thus providing a more robust analysis by mitigating the influence of outliers and fluctuations. We then extracted the relative proportion of athletes showing an increase in EMG activity between $V_{\min}$ and $V_{\max}$, and the proportion of athletes showing a decrease in normalized EMG activity achieved at $V_{\max}$ with increasing resistance. An exact binomial test (two-tailed test) was used to compare each proportion with a theoretical proportion (i.e., 50%).

Changes in the distribution of EMG activity between hamstring muscles. We used the same variables described previously, that is, EMG activity achieved at $V_{\max}$ minus EMG activity achieved at $V_{\min}$ and the difference in normalized EMG activity achieved at $V_{\max}$ between the resistances. These were statistically assessed in using a linear mixed model from the lmerTest package. For the velocity effect, muscle (i.e., ST, SM, BFlh, and BFsh) and resistance condition (i.e., R0, R25, R75) were selected as fixed effects and the intercept of the participants as random effect. For the resistance effect, muscle and resistance comparison (i.e., R25 vs R0, R75 vs R0, R75 vs R25) were selected as fixed effects. Then an ANOVA (degrees of freedom estimated with Satterthwaite formula) was performed on each model. If a main or an interaction effect was found, Holm's *post-hoc* test was used to perform pairwise comparisons.

The alpha threshold was fixed at 0.05. All results are presented as estimated mean $\pm$ standard error.

RESULTS

EMG Activity Modeling as a Function of Normalized Speed

For most of the explored muscles, the second-degree polynomial model showed the highest goodness of fit (hamstrings: mean R^2 stance phase = 0.54, R^2 swing phase = 0.33; gluteals: R^2 stance phase = 0.36, R^2 swing phase = 0.33; quadriceps: R^2 stance phase = 0.39, R^2 swing phase = 0.43; Fig. 2). After careful control of data quality, we considered an average of 10 ± 3 , 14 ± 4 , and 13 ± 4 running cycles by participant for the early acceleration, late acceleration, and maximum speed phases, respectively, for further analysis, including both lower limbs and the two trials of the R0 condition. A total of 11 ± 2 and 9 ± 3 running cycles in the acceleration and maximum speed phases,

respectively, were considered for R25 condition. In the R75 condition, we analyzed 9 ± 3 and 14 ± 7 running cycles in the acceleration and maximum speed phases, respectively. Because of the rapid increase in horizontal velocity during the first steps, the normalized velocity showed an asymmetric distribution with low data density at low velocity values (i.e., <75% of the maximal velocity) and higher data density above this velocity threshold (median = 0.93, Q1 = 0.78, Q3 = 0.98, skewness coefficient = -1.4, kurtosis coefficient = 1.1; Fig. 2). It is worth noting that, after a thorough selection of the generalized models based on Akaike information criterion criteria, the sport factor was significant for numerous muscles, contributing enough to improve the model to warrant its inclusion. However, since the observed differences between track and field athletes and bobsleigh or rugby sevens athletes were relatively minor, and given the imbalance in the proportion of male and female athletes across sports groups, we did not provide a detailed analysis of these results in this study.

During the stance phase, normalized EMG activity of hamstrings increased significantly with increasing velocity, particularly in the unresisted sprint condition (+17.9%; $P < 0.001$). Regardless of the resistance condition, the quadriceps muscles exhibited maximum EMG activity during the acceleration phase. Specifically, the RF and VL reached their maximum EMG activity at 48% and 58% of $V_{\max}$, respectively, in the unresisted sprint condition. GMax showed a linear decrease in EMG activity with increasing velocity across all resisted conditions ($P < 0.05$). In contrast, GMed activity increased up to 85% of $V_{\max}$ in the unresisted sprint condition ($P < 0.001$).

In the swing phase, EMG activity of both the hamstrings and quadriceps exhibited smaller variations with velocity and resistance compared with the stance phase. For all resisted conditions, SM activity peaked at an average of 85% of $V_{\max}$. In the unresisted sprint condition, BFlh showed a polynomial relationship with velocity ($P < 0.001$), with minimum EMG activity occurring at 50% of $V_{\max}$. ST displayed constant EMG activity (32% of maximum EMG activity), whereas the BFsh activity decreased with increasing velocity (-5% of normalized EMG activity, $P < 0.01$). Regardless of the resisted condition, both gluteal muscles showed an overall increase in EMG activity with increasing velocity (+7% in the unresisted sprint condition).

Influence of Stride Cycle on EMG Activity as a Function of Resistance and Sprint Phase

Regardless of the resisted condition or muscle, the mean inter-trial RMSE was 0.0893 ± 0.0244 of normalized EMG activity (range = 0.04–0.17). For each muscle, RMSE was lower than the interindividual variability (0.14 ± 0.03 , range = 0.08–0.21; Table 1).

Unresisted sprint condition. A main effect of the sprint phase ($P < 0.05$) was found on EMG activity for each muscle and each running cycle phase (i.e., stance and swing) except for the RF in the swing phase (Fig. 3).

For most of the stance phase (from 2% to 68% phase duration), EMG activity of all hamstring muscles was higher

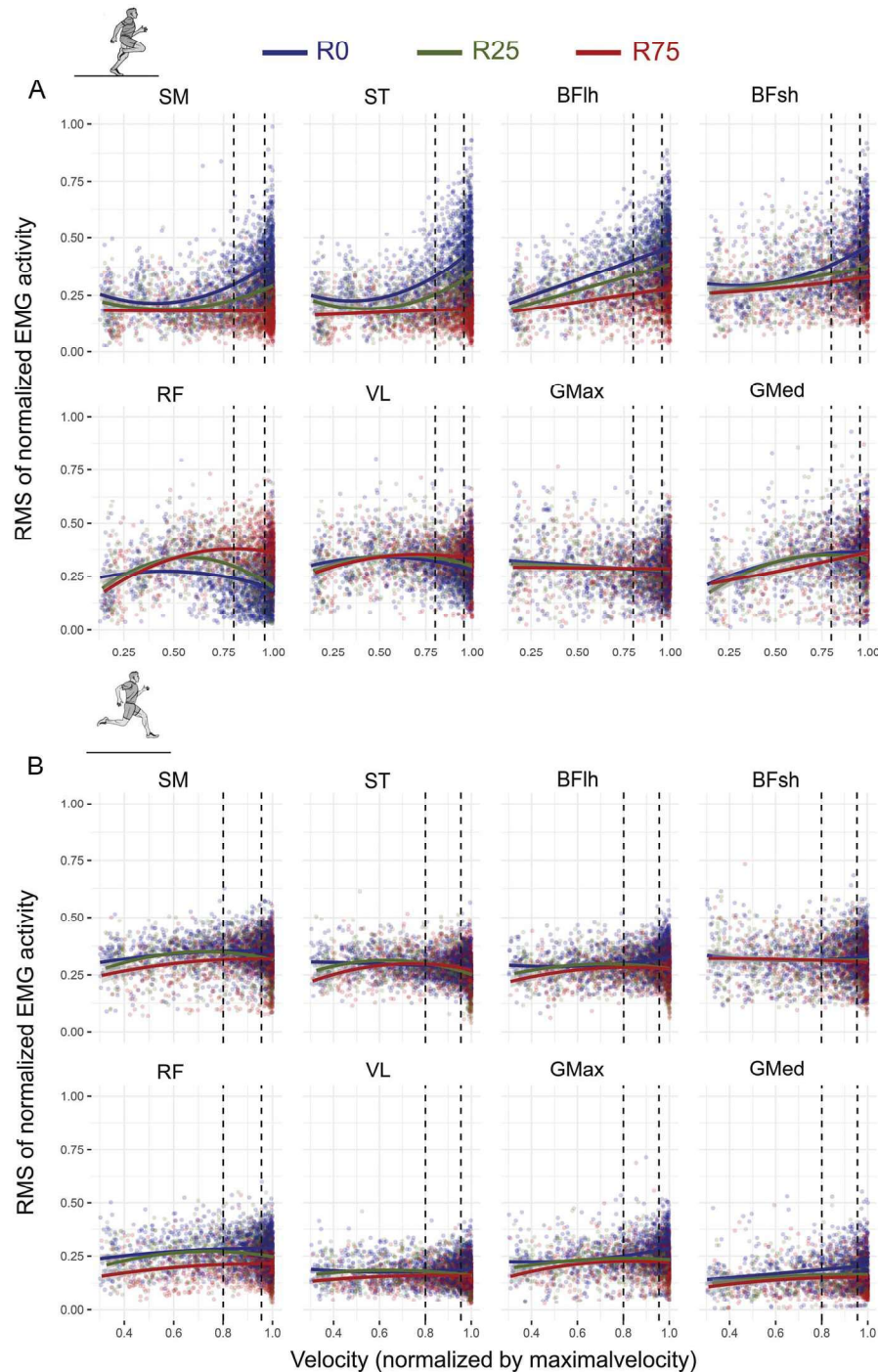


FIGURE 2—Effects of velocity and resisted condition on the RMS of normalized EMG activity for eight lower limb muscles from both lower limbs: (A) during the stance and (B) during the swing running cycle phases. The velocity is normalized by maximum sprint velocity for each condition. The blue, green, and red curves depict the results from the generalized mixed models for the R0, R25, and R75 conditions, respectively. Individual data points represent all steps from each participant. The black vertical dashed lines indicate the average normalized speed at the transition points between sprint phases in the R0 condition (i.e., early acceleration, late acceleration, and maximum speed).

during the maximum speed and late acceleration phases compared with the early acceleration phase ($P < 0.001$). EMG activity was higher in the maximum speed phase as compared with the late acceleration phase for specific parts of the stance phase (ST: 15%–46%, SM: 7%–78%; BFih: 65%–79%, BFsh: 37%–46% and 58%–77% of the stance; $P < 0.01$). During the

early swing, all hamstrings showed a lower EMG activity in the maximum speed phase compared with the early acceleration phase ($P < 0.01$) for relatively short bursts (ST: 1%–22%, SM: 3%–11%, BFih: 5%–9%, BFsh: 3%–8% of the swing). Similar differences were observed between the late acceleration phase and the early acceleration phase for both

TABLE 1. Intertrial RMSE of normalized EMG activity for each muscle, compared with inter-individual variability quantified using standard deviation.

Muscle	Intertrial RMSE	Interindividual Variability
SM	0.09	0.12
ST	0.09	0.13
BFlh	0.09	0.13
BFsh	0.10	0.14
RF	0.09	0.15
VL	0.08	0.15
GMax	0.10	0.17
GMed	0.10	0.14

the ST (1%–23%, $P < 0.001$) and SM (5%–7%, $P < 0.01$). Of the hamstring muscles, only the EMG activity of the BFlh was higher during the late swing phase in the maximum speed

(87%–101%) and late acceleration (93%–101%) phases as compared with early acceleration ($P < 0.001$).

During the stance phase, lower EMG activity was found in the maximum speed phase for RF (11%–60% of the stance phase duration) and VL (34%–60% of the stance, $P < 0.001$) and late acceleration (RF: 11%–66%, VL: 29%–65% of the stance; $P < 0.001$) phases compared with the early acceleration phase. Lower EMG activity was observed in the VL during early acceleration compared with other sprint phases at 62%–71% of the swing phase ($P < 0.001$), whereas it was higher at 80%–91% of the sprint phase ($P < 0.001$). No differences were observed in the RF in the swing phase.

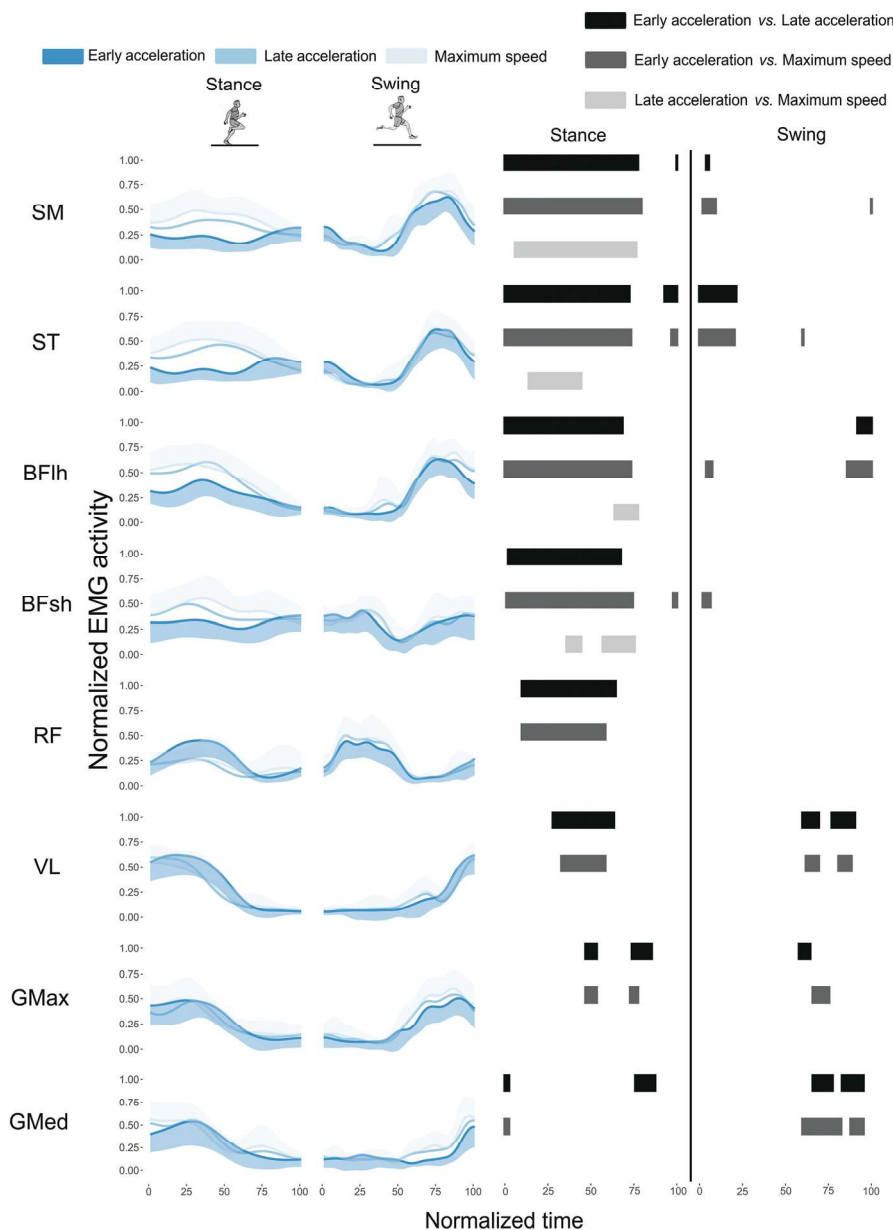


FIGURE 3—Effect of the sprint phase on the EMG activity of eight lower-limb muscles from the rear leg at the start of the sprint. The left panel represents the group mean $\pm$ standard deviation of normalized EMG activity during the early acceleration, late acceleration, and maximum speed phases of the R0 condition. Note that to make the graph easier to read, the standard deviation is shown only for the maximum speed and early acceleration phases in one direction only. The curves are shown for both running cycle phases (stance or swing) with a normalized time. The right panel represents the SPM results of the comparisons between the sprint phases (a colored rectangle appears when $P < 0.05$ for each normalized time step).

Concerning the gluteal muscles, only GMed showed lower EMG activity in the early acceleration phase compared with the late acceleration (1%–4%, $P < 0.05$) and maximum speed (1%–4%, $P < 0.05$) phases during the early stance. During the late stance, both gluteal muscles had a lower EMG activity in the early acceleration compared with the late acceleration phase (GMax: 75%–87%, $P < 0.01$; GMed: 77%–89%, $P < 0.01$) and only for GMax compared with the maximum speed phase (74%–79%, $P < 0.01$). In the swing phase, both GMax and GMed were less activated in the early acceleration phase as compared with late acceleration (GMax: 59%–66%, GMed: 67%–79% and 84–97%; $P < 0.001$) and maximum speed (GMax: 67%–77%, GMed: 61%–84% and 89–97%; $P < 0.001$) phases.

Comparison of resistance conditions for each sprint phase. EMG activity showed a main effect of the sprint phase and resistance for all muscles ($P < 0.05$; Figs. 4 and 5). We found a resistance–sprint phase interaction for all muscles in each running cycle phase excepted for the BFsh in swing phase.

For most of the stance phase and regardless of the sprint phase, all hamstring muscles, except for BFsh, presented a lower EMG activity at higher resistance conditions ($P < 0.001$). BFsh displayed a higher EMG activity in the R0 condition compared with R25 and R75 during the second half of the stance phase in both acceleration (R0–R25: 61%–86% of the stance, R0–R75: 56%–78% of the stance; $P < 0.001$) and maximum speed phases (R0–R25: 61%–88% of the stance, R0–R75: 45%–79% of the stance; $P < 0.001$). In the swing phase, although we did not observe differences in BFsh between resistance levels, the EMG activity of all the other hamstring muscles decreased with increasing resistance during the late swing phase in the acceleration sprint phase ($P < 0.001$). In the maximum speed phase, SM, ST, and BFlh activity decreased in the R0 condition as compared with both resisted conditions ($P < 0.001$), whereas no difference was found between R25 and R75. In the R75 condition, the SM and ST showed higher EMG activity for short bursts within the early swing phase as compared with the other resistance levels.

Regarding the quadriceps muscles, EMG activity was higher throughout most of the stance phase at higher resistance levels during both the acceleration and maximum speed phases ($P < 0.001$), whereas no difference was found between R0 and R25 during the maximum speed phase. During the middle of the stance phase, VL showed higher EMG activity in both R25 and R75 conditions during the acceleration phase as compared with R0, and in the R75 condition as compared with R25 and R0 during the maximum speed phase ($P < 0.001$). In the early swing phase, RF EMG activity decreased in the R75 condition as compared with the others resistance levels in both the acceleration and maximum speed phases ($P < 0.001$). EMG activity of VL decreased briefly in the second half of the swing phase with increasing resistance (R75 compared with R0: in the acceleration phase: 60%–72% of the swing phase, $P < 0.001$; in the maximum speed phase: 59%–76% of the swing phase, $P < 0.001$).

In the acceleration phase, EMG activity of GMax was higher during the early stance in the R75 condition as compared with R0 condition ($P < 0.01$). Differently, EMG activity

of GMed was lower in the late stance phase in the R75 condition as compared with R0 condition ($P < 0.01$). EMG activity increased in both GMax and GMed in the R0 condition during the second half of the swing phase ($P < 0.01$).

Between-Individual Consistency of the Effects of Velocity and Resistance on EMG Activity

Regarding the hamstring muscles in the stance phase, a high proportion of athletes showed an increased EMG activity (i.e., difference of EMG at $V_{\max}$ minus EMG at $V_{\min}$) with increasing velocity for almost all resistance conditions (range = 70%–100%, $P < 0.05$; Table 2). Only the ST and SM muscles presented no significant change in the R75 condition. This proportion decreased progressively with increasing resistance (e.g., from 97% in R0 to 74% in R75 for the BFlh). For all hamstring muscles heads, almost all athletes showed lower EMG activity at $V_{\max}$ as resistance increased (e.g., proportion equal to 100% for SM and ST depending of the resistance comparison, $P < 0.001$).

In the swing phase, no clear trend emerged for most hamstring muscles (proportion close to 50%). More athletes showed decreases in EMG activity at $V_{\max}$ as compared with the start of the sprint for ST in the R25 condition (proportion of increased EMG activity at $V_{\max} = 22\%$, $P < 0.01$). In contrast, a high proportion of athletes exhibited an increased EMG activity in BFlh at $V_{\max}$ in the R0 condition (73%, $P < 0.01$) and for SM in the R75 condition (69%, $P < 0.05$). Both BFlh and SM presented a high between-individual consistency of lower EMG activity at $V_{\max}$ with increasing resistance ($P < 0.001$). In addition, a significant proportion of participants exhibited lower BFsh and ST EMG activity in the R0 condition compared with the R25 condition ($P < 0.01$), but no significant difference was observed compared with the R75 condition.

For the quadriceps muscles in the stance phase, both VL and RF presented a high proportion of athletes with an increased EMG activity with increasing velocity but only in the R75 condition ($P < 0.01$). Regarding the resistance effect, a significant proportion of athletes showed a higher EMG activity at $V_{\max}$ in the R75 condition compared with R25 ($P < 0.05$) and R0 ($P < 0.001$).

In the swing phase, a significant proportion of athletes showed a decreased VL EMG activity with increasing velocity but only in the R25 condition ($P < 0.001$). Both VL and RF showed a significant proportion of athletes with increasing EMG activity at $V_{\max}$ in the R0 condition as compared with R25 ($P < 0.001$) and R75 ($P < 0.05$).

In the gluteal muscles, no significant proportion was observed for the GMax in the stance phase, whereas a high proportion of athletes presented an increase in the EMG activity of GMed with increasing velocity for all resistance conditions (range = 81%–96%, $P < 0.01$). Regarding the resistance effect, only GMax showed a significant proportion of athletes with increasing EMG activity at $V_{\max}$ in the R75 condition compared with R0 ($P < 0.05$).

In the swing phase, both GMax and GMed showed a significant proportion of athletes with positive values in the difference

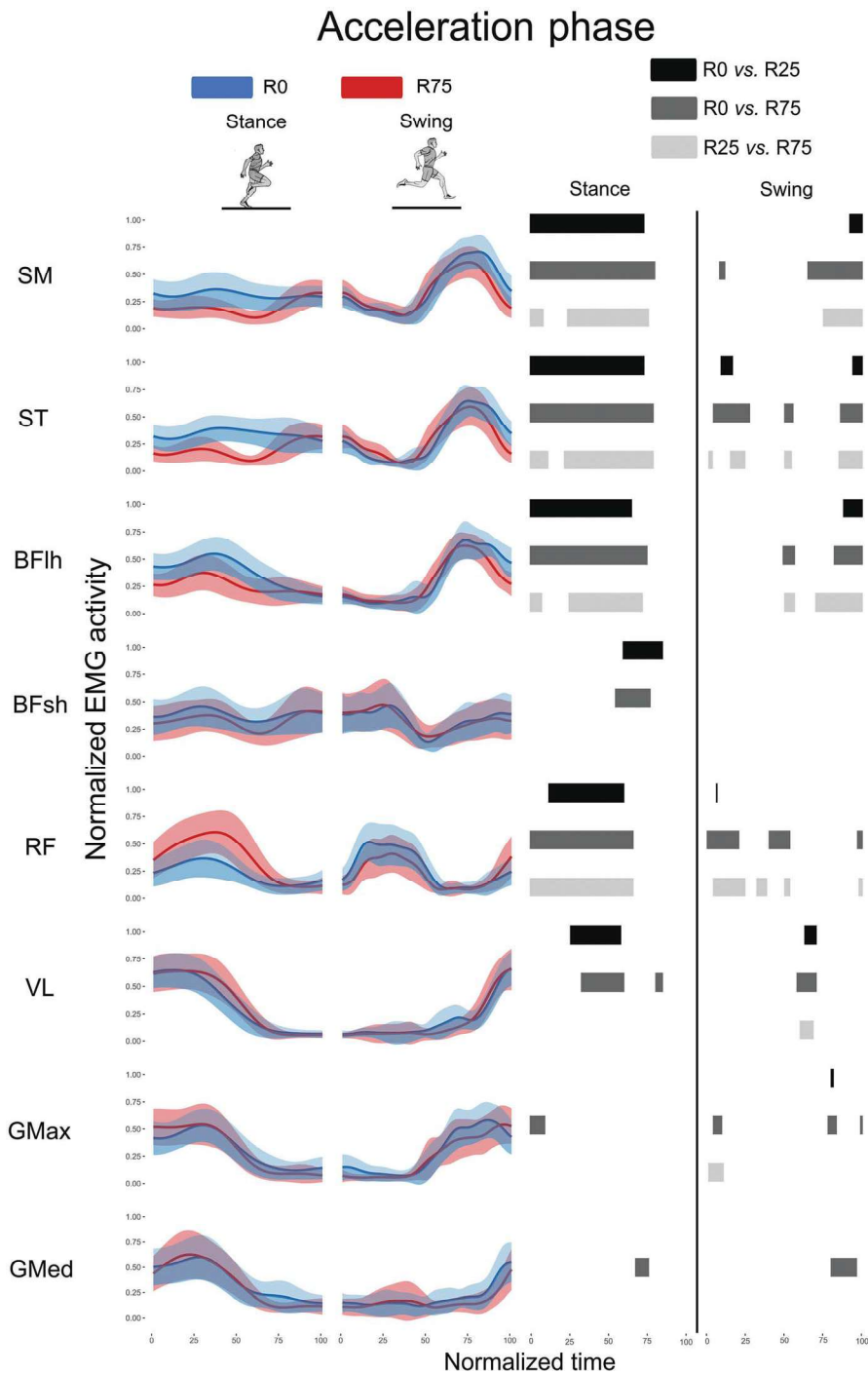


FIGURE 4—Effect of the resisted condition on the EMG activity of eight lower-limb muscles from the rear leg at the start of the sprint. The left panel represents the group mean $\pm$ standard deviation of normalized EMG activity during the acceleration phase of the R0 (blue) and R75 (red) conditions. The curves are shown for both running cycle phases (stance or swing) with a normalized time. To simplify the visualization, the R25 condition is not shown but appears in the SPM results. The right panel represents the SPM results of the comparisons between the resisted conditions (a colored rectangle appears when $P < 0.05$ for each normalized time step).

of EMG at $V_{\max}$ minus EMG at $V_{\min}$, but only in the R0 and R75 conditions ($P < 0.05$). Both GMax and GMed showed a high proportion of athletes presenting a decrease EMG activity at $V_{\max}$ with increasing resistance (range = 80%–86%, $P < 0.05$).

Difference between Hamstring Muscle Heads in EMG Activity Changes

We found a main effect of muscle ($P < 0.001$) and resisted condition ($P < 0.001$), with no muscle–condition interaction

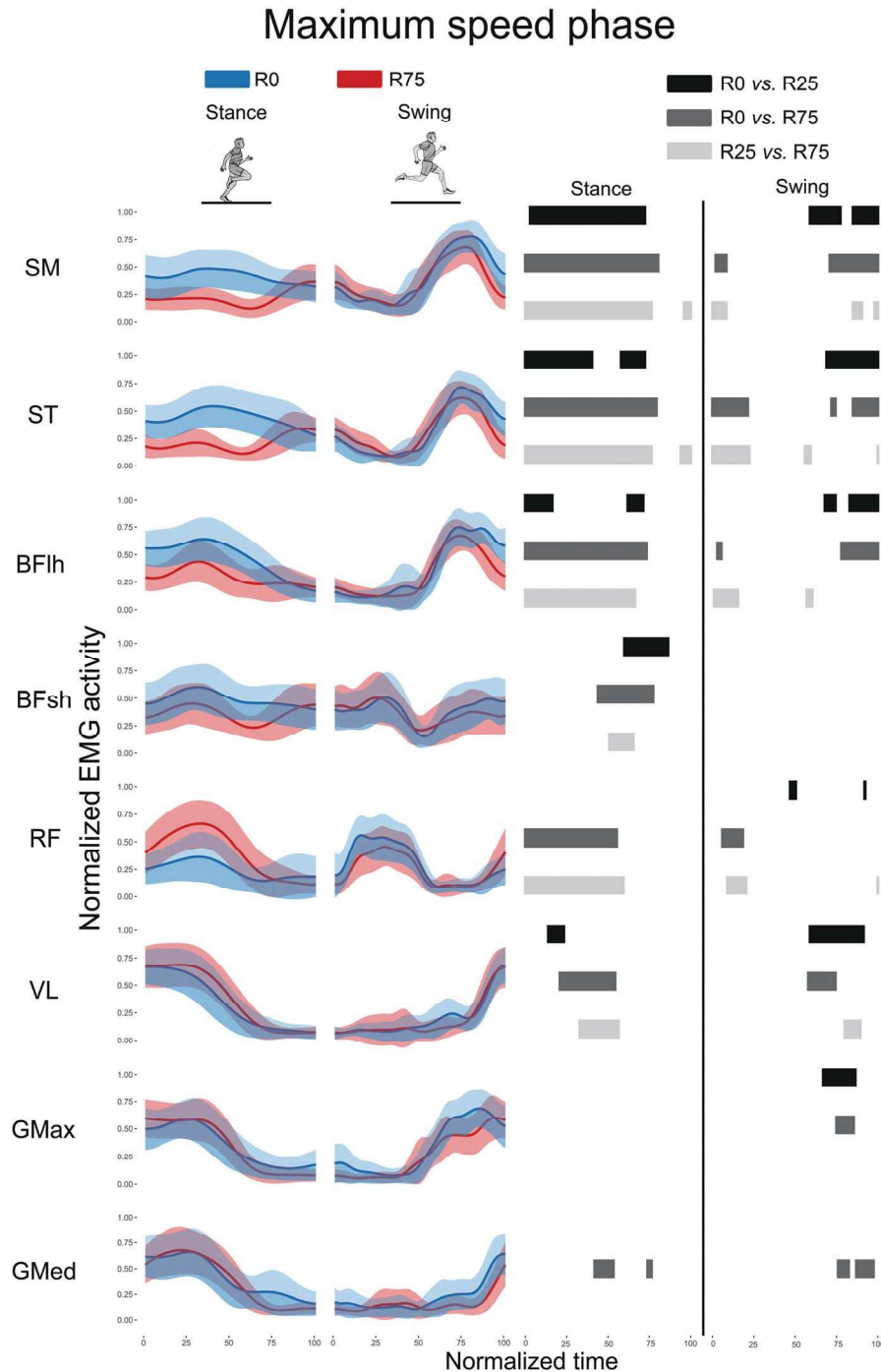


FIGURE 5—Effect of the resisted condition on the EMG activity of eight lower-limb muscles from the rear leg at the start of the sprint. The left panel represents the group mean $\pm$ standard deviation of normalized EMG activity during the maximum speed phase of the R0 (blue) and R75 (red) conditions. The curves are shown for both running cycle phases (stance or swing) with a normalized time. To simplify the visualization, the R25 condition is not shown but appears in the SPM results. The right panel represents the SPM results of the comparisons between the resisted conditions (a colored rectangle appears when $P < 0.05$ for each normalized time step).

on EMG at $V_{\max}$ – EMG at $V_{\min}$ (Fig. 6A) for both stance and swing phases. Holm's *post hoc* revealed that independently of the muscle, the difference between EMG activity at $V_{\max}$ and EMG activity at $V_{\min}$ was the highest in the R0 condition for both the stance and swing phases ($P < 0.001$), and higher in the R25 condition as compared with the R75 in the stance phase ($P < 0.001$) but lower in the swing phase ($P < 0.001$). Regardless

of the resistance condition, in the stance phase, BFllh showed a higher difference in EMG activity between $V_{\max}$ and $V_{\min}$ as compared with SM (BFllh – SM = $6.5\% \pm 1.3\%$, $P < 0.001$), BFsh (BFllh – BFsh = $4.2\% \pm 1.3\%$, $P < 0.01$), and ST (BFllh – ST = $3.2\% \pm 1.3\%$, $P < 0.05$). ST showed a higher difference in EMG activity between $V_{\max}$ and $V_{\min}$ as compared with SM (ST – SM = $3.3\% \pm 1.3\%$, $P < 0.05$). In the swing phase, BFllh

TABLE 2. Proportion of athletes showing (i) an increase in EMG activity between $V_{\min}$ and $V_{\max}$ (i.e., EMG at $V_{\max}$ – EMG at $V_{\min}$ (%)) for each resistance condition, and (ii) a decrease in normalized EMG activity achieved at $V_{\max}$ with increasing resistance (i.e., difference of EMG at $V_{\max}$ between resisted conditions (%)) for each condition comparison.

Muscle	EMG at $V_{\max}$ – EMG at $V_{\min}$ (%)			Difference of EMG at $V_{\max}$ between Resisted Conditions (%)		
	R0	R25	R75	R0–R25	R0–R75	R25–R75
Stance	ST	95	94	66	92	100
	SM	95	89	60	97	100
	BFlh	100	92	74	83	97
	BFsh	97	92	70	83	91
	VL	54	62	76	42	21
	RF	61	51	74	54	18
	Gmax	34	52	50	43	36
	GMed	86	81	96	65	60
Swing	ST	51	22	49	92	69
	SM	65	33	69	97	86
	BFlh	73	36	43	97	89
	BFsh	54	36	45	75	58
	VL	63	21	45	91	73
	RF	67	49	62	83	79
	Gmax	94	55	71	83	86
	GMed	71	64	72	76	80

Proportions significantly different from 50% are shown in bold according to an exact binomial test.

and SM were the only hamstring muscles to show an increase in EMG activity at $V_{\max}$ compared with EMG activity at $V_{\min}$ ($P < 0.01$). In contrast, both ST and BFsh showed a decrease in EMG activity and did not differ significantly from each other.

We observed a main effect of muscle ($P < 0.001$) and condition comparison ($P < 0.001$) on EMG activity at $V_{\max}$, whereas no significant muscle–resistance comparison interaction was found ($P = 0.086$) for both stance and swing phases (Fig. 6B).

In the stance phase, independently of the muscle, the difference in EMG activity at $V_{\max}$ was the highest in the R0 versus R75 comparison compared with R25 versus R75 and R0 versus R25 ($+10.9\% \pm 1.1\%$, $P < 0.001$). In the swing phase, only the R25 versus R75 showed a significant positive difference ($1.5\% \pm 0.6\%$) compared with the other comparisons ($P < 0.001$), which presented negative differences ($R0-R25 = -6.4\% \pm 0.6\%$; $R0-R75 = -4.8\% \pm 0.6\%$). Regardless of the condition comparison, in the stance phase, ST presented higher differences in EMG activity at $V_{\max}$ between the resistance conditions than BFlh ($BFlh - ST = 5\% \pm 1.3\%$, $P < 0.001$) and BFsh ($BFlh - ST = 6.9\% \pm 1.3\%$, $P < 0.001$), whereas no difference was found when compared with SM. SM showed a larger difference in EMG activity as compared with BFsh ($BFsh - SM = 4.8\% \pm 1.3\%$, $P < 0.01$) but was not different from BFlh. In the swing phase, the BFlh showed lower differences in EMG activity at $V_{\max}$ between the resisted conditions than ST ($BFlh - ST = -3\% \pm 0.8\%$, $P < 0.001$) and BFsh ($BFlh - BFsh = -3.4\% \pm 0.8\%$, $P < 0.001$). The difference was lower in the SM as compared with BFsh ($SM - BFsh = -2.2\% \pm 0.8\%$, $P < 0.05$).

DISCUSSION

The present study provides a detailed comparison of the effect of running speed and resistance on the EMG activity of the gluteal and thigh (i.e., hamstrings and quadriceps) muscles.

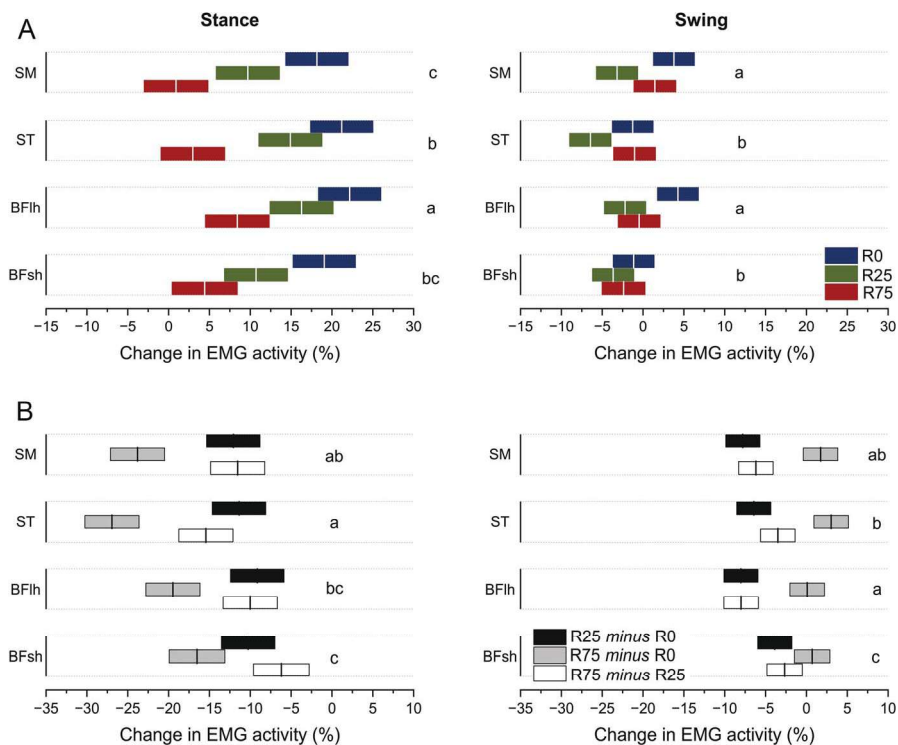


FIGURE 6—Velocity and resisted condition effects as function of the hamstring muscle head. A, Differences (in %) between normalized EMG activity achieved at $V_{\max}$ minus the normalized EMG activity at minimal velocity for both stance and swing running cycle phases. B, Comparisons of the normalized EMG activity achieved at $V_{\max}$ between the resisted conditions. Bar plots sharing the same letter (a, b, c, d) are not to be considered significantly different according to the main effect of muscle in linear mixed models after *post-hoc* analysis.

The lower reproducibility error compared with the interindividual variability of the observed EMG signals underscores the robustness of the present data. We are therefore confident that the present results provide a reliable examination of the effect of velocity and resistance on EMG activity. Firstly, we found that EMG activity increased in the hamstring and decreased in the quadriceps with increasing running velocity. Conversely, hamstrings EMG activity decreased while quadriceps activity increased with increasing resistance. These changes were specific to distinct phases of the running cycle and sprint phase (Figs. 2–5), with high variations when muscles were highly activated within the sprint cycle, that is, at the start of the stance phase and in the late swing phase for the hamstrings. Second, velocity and resistance showed consistent increases in the EMG activity of hamstrings for 90% of the participants, particularly in the stance phase. Third, the hamstring muscle heads were differently affected by velocity and resistance: the BFLh muscle exhibited the highest increase in EMG activity as velocity increased, whereas the ST muscle showed the greatest decrease as resistance increased.

Velocity and resistance effects. This study is the first to model EMG activity in relation to speed during sprint runs over 40 m using a nonlinear approach. Polynomial modeling revealed that hamstring muscles' EMG activity increased with increasing velocity, whereas it inversely decreased with increasing resistance. Opposite effects were observed in the quadriceps muscles (Fig. 2). The largest increase in EMG activity throughout the sprint acceleration phase was observed in the hamstring muscles, as compared with the quadriceps and gluteal muscles. These findings underline the large contribution of the hamstrings to hip extension velocity, propulsive impulse, and in turn maximum running speed (7,27). Of note, the increase in hamstring EMG activity during the stance phase with increasing velocity and decreasing resistance was highly consistent across individuals. For example, 100% of the participants showed a decrease in SM and ST EMG activity in the highest resistance condition compared with the unresisted condition. The opposite changes between quadriceps and hamstring muscles could potentially originate from their different anatomical position (28). Furthermore, hamstrings muscle fascicles may operate on a favorable portion of the force–length relationship at maximum speed. Oppositely, the inclined body position in the acceleration phase (29) is thought to induce an excessive stretching of the hamstring muscles and potentially imposes muscles fascicles to operate in a mechanically unfavorable scenario (18).

The statistical models revealed that the mean EMG changes in the hamstrings with increasing velocity were significantly smaller during the swing compared with the stance phase. The consistency of the velocity effect on the hamstrings EMG activity across individuals was less than 70% for the majority of the observed proportions. This may be explained by the fact that averaged EMG activity could conceal short-duration changes occurring in the swing phase. Furthermore, the time-series analysis confirmed that the changes in the hamstring muscle activity mainly occurred during their most activated periods, specifically in the first half of the stance and late swing phase (30).

During the late swing phase, EMG activity decreased in all bi-articular hamstring muscle heads (i.e., BFLh, SM, and ST) with additional resistance, whereas EMG activity increased only in the BFLh with increasing velocity (in the unresisted sprint condition). This finding suggests that resistance has a more predominant effect than velocity on hamstring EMG activity. The fact that only the BFLh increased its EMG activity with increasing velocity during the late swing phase, where high muscle fiber strain is sustained (31), could imply a greater contribution to the knee flexion and hip extension torque and partly explain the higher incidence of strain injuries affecting the BFLh muscle in sprint-based sports (32). However, the difference in BFLh EMG activity between maximum speed and early acceleration was substantially smaller than the differences one would expect from previous studies that examined BFLh EMG activity during different speeds of steady-speed running (4,8). This discrepancy may be due to the fact that most previous studies focused on sprints at steady speed, whereas the present study explored *in situ* sprints with acceleration. Although the reasons for the reduced changes in EMG activity during acceleration running are not entirely clear, recent studies have highlighted that hamstring muscle–tendon mechanics differ between acceleration and steady-state running at matched speeds. Specifically, MTU length of the BFLh is greater and lengthens more rapidly during acceleration due to increased pelvis tilt and hip flexion angles, despite greater knee flexion (33,34).

The time-series analysis revealed that bi-articular hamstring muscles were less activated in the early swing phase with decreasing resistance and increasing velocity. This could be explained by the need to reduce co-activation of antagonist muscles (i.e., between hamstring and RF). This reduction in co-activation of hip extensors may likely facilitate rapid hip flexion at high velocities. We cannot exclude that reflex mechanisms (35), such as reciprocal inhibition, contribute to reduced excitation in the biarticular hamstring muscles. This observation contrasts with the first half of the stance phase, where peak activation of the RF and hamstrings appears to be in phase, regardless of the resistance condition. Such muscle coordination could enhance knee joint stabilization or increase force production, as the hamstrings act as agonists to the quadriceps in a closed chain, promoting knee extension (36).

Low consistency was observed in the effect of increased quadriceps EMG activity with increasing resistance and decreasing velocity, although approximately 80% of individuals exhibited an increase in RF or VL EMG activity during the stance phase in the R75 condition compared with R0. This may be explained by the fact that maximum average EMG activity in quadriceps is not obtained at $V_{\max}$ or $V_{\min}$, as revealed by the polynomial models. Furthermore, the rapid increase in average EMG activity of both VL and RF with increasing velocity during the stance phase (reaching maximum EMG activity at approximately 50% of $V_{\max}$) may be explained by the rapid increase in braking force during the initial steps of sprint acceleration (37). Furthermore, Pandey et al. (38) reported that the primary function of VL and RF in sprinting is to support the body in the stance phase through their large

contribution to knee extensor torque, and contribute to leg stiffness and vertical forces, reflecting the energy absorber action of knee joint (39). However, except for the highest tested resisted condition (R75), modeling of mean EMG activity as a function of speed (Fig. 2) revealed that both VL and RF reached their maximal EMG activity before achieving maximal velocity during the early acceleration phase, and then slightly decreased their EMG activity. Thus, the quadriceps muscles are assumed to be highly activated in the acceleration phase in order to produce propulsive force in inclined body position (40). Although this hypothesis conflicts with the braking role of the quadriceps reported by Pandey et al. (38), their synergistic action with hamstrings and the triceps surae could stabilize the knee joint, favor the transfer of energy to assist plantar flexor torque generation and, in turn, enable an effective push-off at the ankle joint (41).

According to their distinct anatomical function, specific changes were found between RF and VL in the timing of their peak EMG activity, occurring at 48% of $V_{\max}$ for RF and 58% of $V_{\max}$ for VL. According to the time-series analysis, only VL presented an increase EMG activity at the beginning of the second half of the swing phase in the maximum speed phase compared with other sprint phases as well as in the unresisted sprint condition compared with resisted conditions. This observation is in agreement with previous studies (42–44) and could at least partly be explained by the major function of the VL, that is, to extend the knee during late swing. Furthermore, the front leg extension amplitude has been reported to be higher with increasing velocity (44). As previously mentioned, only RF showed an increased EMG activity during the early swing with decreasing resistance in both acceleration and maximum speed phases, which could be explained by its bi-articular function enabling rapid hip flexion during high velocity movements.

Although GMax and GMed are both hip extensors, their roles in sprinting appear to differ significantly (38) in accordance with their specific architectures. As running velocity increased, the average GMax EMG activity showed minimal variations during the stance phase, whereas it increased during the swing phase. Contrastingly, the average GMed EMG activity presented minimal variations during the swing phase and increased in the stance phase. These changes align with previous studies suggesting that the primary role of GMax is to assist the hamstrings in late swing, as it does not substantially contribute to forward or vertical acceleration during the stance phase (38,45). However, the analyses of average EMG activity may conceal specific changes occurring within the running cycle. In this sense, the time-series analysis showed an increase in both GMax and GMed EMG activity in the second half of the swing phase with increasing velocity or decreasing resistance. Thus, all gluteal muscles could assist the hamstrings in increasing the hip extension torque during the swing phase. Emphasizing the major role of GMax during sprint running, elite sprinters have relatively larger GMax volumes compared with untrained individuals and sub-elite athletes (46). Although the role of the GMed during sprint running remains poorly understood, GMed volume is not suggested as a critical determinant of sprint performance (47,48). However,

Bartlett et al. (49) reported significantly higher GMed EMG activity during sprinting compared with walking, climbing, or running, emphasizing its critical role in sprint performance. Interestingly, GMed demonstrated a brief yet significant increase in EMG activity at the onset of stance with increasing velocity. This is in line with previous studies showing higher GMed EMG amplitude as running speed increases (50). Considering that the contribution of gluteal muscles to propulsion during stance is relatively small compared with other hip extensors (38), this increased activation of GMed could contribute to compensate involuntary movements elicited by high velocity running to stabilize the pelvis and hip at high speeds (51,52). The present results also align with the high increase in force produced by the GMed between the first and last sprint foot contact reported by Pandey et al. (38). A key and unexpected finding of this study was the increase in GMed EMG activity during the late swing phase, both with higher velocity and reduced resistance. Similar to the stance phase, this increase may reflect the need to pre-activate the muscle to generate force before foot ground contact. An alternative explanation could be the influence of step frequency, which increases during sprinting and is higher in unresisted sprints. Supporting this, Chumanov et al. (6) observed that GMed EMG activity rises in the late swing phase as step frequency increases during constant-speed running.

Effects of velocity and resistance on the different hamstring heads. The comparison of the velocity and resistance effects between hamstring muscle heads (Fig. 6) revealed that the distribution of EMG activity changes at maximal velocity compared with the start of the sprint. During the stance phase, EMG activity increased the most in BFLh from sprint start to $V_{\max}$, followed by ST, BFsh, and SM. When comparing ST and SM, it is likely that the relative changes in EMG activity from early acceleration to $V_{\max}$ may reflect the distinct architecture of these muscles. Furthermore, ST has long fascicles and more compliant aponeurosis (17), which are advantageous at high contraction velocities (53). In this sense, increasing resistance (which ultimately decreases running speed) resulted in the highest decrease in ST EMG activity during the stance phase, which can still be explained by its quasi-fusiform architecture that is less suitable for high force production than pennate muscles. This is in accordance with the selective hypertrophy of the ST in sprinters compared with non-sprinters (47) or in elite sprinters compared with sub-elite sprinters (46). Interestingly, the least pronounced decrease in EMG activity at $V_{\max}$ in the stance phase with increasing resistance was observed in BFsh and BFLh. This is somewhat inconsistent with the observed changes with increasing running velocity, which was unexpected. Although BFsh also has long fascicles compared with SM and BFLh, its contribution to propulsion is likely minimal due to its mono-articular function. It should be noted that this muscle has rarely been studied before, and time-series analysis clearly shows that this muscle head has a very different role from the other hamstrings. The BFsh is also active during the early swing phase, likely contributing to maximum knee flexion to reduce the leg's moment of inertia, thereby enabling a higher lower leg velocity (54).

In the late swing, time-series analysis revealed that only the BFlh EMG activity increased in the maximum speed phase as compared with the acceleration phase. Because BFlh MTU lengthens the most in late swing phase compared with other hamstring MTU (55,56), this EMG activity increase may be an efficient strategy to protect BFlh muscle fibers from being stretched beyond tolerable limits. Furthermore, given that the BFlh is the most frequently injured muscle in sprint-based sports (57), the present findings highlight the importance of achieving high running speeds during sprint training as a key prevention method to enhance the recruitment of this muscle (58).

Limitations. Because of the demanding experimental protocol and the number of sensors used (16 sensors for 8 muscles in both lower limbs), some muscles involved in sprint running have not been investigated in this study. For example, calf muscles play a key role in accelerating the body forward across the acceleration phase (38,59–61), and iliopsoas has a large contribution to hip flexion in the swing phase (38) as well as to the high hip- and knee-joint accelerations observed at high running speeds (7). The protocol included only two resisted conditions, with the highest (R75) inducing approximately a 50% decrease in maximum velocity. Future studies should explore the evolution of EMG activity in more extreme regions of the load-velocity relationship, either by using higher resistance conditions or by investigating overspeed condition in assisted sprinting. It should be noted that, due to the minor differences observed between the front leg and rear leg, the stride cycle analysis using SPM was focused exclusively on the rear leg. Future investigations should further explore the differences between the two legs, particularly during the sprint start where velocity condition changes rapidly between steps. Because we used surface electrodes, EMG signals might have been contaminated by cross-talk. In this context, we used ultrasonography to ensure we place the electrodes as far from the muscle borders as possible, and we also used 10-mm inter-electrode distance in double-differential mode to minimize cross-talk. It should also be mentioned that bipolar EMG used in this study may not reflect the EMG activity of the entire muscle (62). However, the use of the spatially robust high-density EMG in such a large number of muscles during sprinting has technical limitations. It should also be mentioned that the changes in muscle geometry during the sprint acceleration might have affected EMG amplitudes (63); therefore, direct conclusions on neural drive to the muscles should be drawn with caution. One may note that a crossover design would allow to check whether the order of the conditions could have influenced the results, particularly due to potential post-activation potentiation effects or other subcutaneous changes (e.g., sweat or increased muscle temperature). The present experimental design ensured a proper warm-up, while limiting potential fatigue occurrence and ensuring progressive

increase in resistance. Such approach was also crucial to optimize the adherence of elite athletes. Finally, resistance conditions were compared in terms of normalized speed values as well as sprint phases. This implies that the conditions were not compared at the same absolute speed, which must be considered when interpreting the present results.

Practical implications. The observation that increasing resistance produces the opposite effect to increasing velocity on the quadriceps and hamstring muscles suggests that resisted sprints could potentially serve as an effective training method to mimic the early phases of sprint acceleration. This approach may specifically stimulate adaptations in the VL, in contrast to unresisted sprinting, which might primarily target adaptations in the hamstrings (particularly the BFlh and ST). However, this hypothesis requires validation through future randomized controlled training interventions. Because hamstrings were relatively less activated when resistance increased, resisted sprinting may be introduced earlier during rehabilitation after a hamstring injury than maximum unresisted sprint. This study also warrants future research distinguishing between hamstring muscles' contribution to sprinting because of the muscle-specific response to increasing resistance and running speed. Another noteworthy observation from our study is the important role that GMed seems to play as velocity increases, probably because of its role as a stabilizer of the pelvis. Sprint training should therefore include exercises specifically aimed at recruiting postural muscles including GMed. Further studies should explore the synergies between muscles during sprinting in order to better understand the contribution of these postural muscles to sprint running. This could potentially help to detect coordination patterns that are beneficial to performance. This could potentially help to detect coordination patterns that are beneficial to performance. Finally, the analysis of between-individual consistency regarding the effects of velocity and resistance revealed some inter-individual variabilities. This underscores the importance of individualized assessment to effectively apply the findings from this study and for better individualizing both injury prevention and sprint performance.

The authors warmly thank the athletes who participated in the FULGUR project. The present study was conducted in the context of the FULGUR project (ANR-19-STPH-003) funded by the French Research Agency in the perspective of the Paris 2024 Olympic and Paralympic Games in collaboration with French Federations of Athletics, Rugby and Ice Sports, Universities of Nantes, Côte d'Azur, Savoie Mont Blanc, Jean Monnet Saint-Etienne, Saclay, the Mines Saint-Etienne, the CEA, and the CNRS. The authors report no conflicts of interest, financial or otherwise. The results of the study are presented clearly, honestly, and without fabrication, falsification, or inappropriate data manipulation. The results of the present study do not constitute endorsement by the American College of Sports Medicine.

Data Availability Statement: Data are available upon reasonable request. Requests for data sharing from appropriate researchers and entities will be considered on a case-by-case basis. Interested parties should contact the corresponding author Robin Macchi (robin.macchi@gmail.com).

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