POLITECNICO DI TORINO
Repository ISTITUZIONALE

On the Detection of High-Quality, High-Density Electromyograms During 80m Sprints: a Case Study

Original
On the Detection of High-Quality, High-Density Electromyograms During 80m Sprints: a Case Study / Nicola, Riccardo; Cerone, GIACINTO LUIGI; Caruso, Marco; Rossanigo, Rachele; Cereatti, Andrea; Martins, Taian. - ELETTRONICO. - (2022), pp. 1-5. (Intervento presentato al convegno IEEE International Workshop on Medical Measurement and Applications (MEMEA) tenutosi a Messina (Italy) nel 22-24 June 2022) [10.1109/MeMeA54994.2022.9856504].

Availability:
This version is available at: 11583/2972225 since: 2022-10-18T10:49:06Z

Publisher:
IEEE

Published
DOI:10.1109/MeMeA54994.2022.9856504

Terms of use:

This article is made available under terms and conditions as specified in the corresponding bibliographic description in the repository

Publisher copyright
IEEE postprint/Author's Accepted Manuscript

©2022 IEEE. Personal use of this material is permitted. Permission from IEEE must be obtained for all other uses, in any current or future media, including reprinting/republishing this material for advertising or promotional purposes, creating new collecting works, for resale or lists, or reuse of any copyrighted component of this work in other works.

(Article begins on next page)

20 April 2024

On the Detection of High-Quality, High-Density Electromyograms During 80m Sprints: a Case Study

Riccardo Nicola
Politecnico di Torino
Turin, Italy
s279924@studenti.polito.it

Giacinto Luigi Cerone
Department of Electronics and Telecommunications
Politecnico di Torino
Turin, Italy
giacintoluigi.cerone@polito.it

Marco Caruso
Department of Electronics and Telecommunications
Politecnico di Torino
Turin, Italy
marco.caruso@polito.it

Rachele Rossanigo
Department of Biomedical Sciences
University of Sassari
Sassari, Italy
r.rossanigo@studenti.uniss.it

Andrea Cereatti
Department of Electronics and Telecommunications
Politecnico di Torino
Turin, Italy
andrea.cereatti@polito.it

Taian Vieira
Department of Electronics and Telecommunications
Politecnico di Torino
Turin, Italy
taian.martins@polito.it

***Abstract*—Surface electromyograms (EMGs) have been often used to study muscle function in locomotor activities. Typically, EMGs are sampled with a single pair of electrodes, providing information on the timing and degree of muscle excitation. Additional information may be obtained when sampling EMGs with multiple electrodes from the same, target muscles. Studies using high-density EMGs (HD-EMGs) in locomotor activities are limited to laboratory settings and low speed tasks, likely due to the technical shortcomings in the commercially available systems for high-density recordings. This issue is further aggravated when kinematics data are necessary for associating EMGs with events of interest during the movement cycle. By combining two systems, ad hoc developed for the on-field recording of kinematics data and HD-EMGs, here we present single-case results during extreme-speed locomotion–the 80 m sprint on an official, athletic track. Our aim was to verify whether descriptors of quality documented in the EMG literature during well-controlled, isometric contractions, apply to the HD-EMGs we detected and segmented with respect to the running cycles. From a single, elite sprinter, we were able to obtain HD-EMGs with negligible movement artifacts and with temporal profiles typically characterizing action potentials of single motor units. Our results would seem to advocate the possibility of using HD-EMG to study muscle function during highly dynamic contractions outside the laboratory settings.**

Keywords—high-density EMG, inertial sensor, running, sprint, gastrocnemius

I. Introduction

The possibility of obtaining information on the timing and degree of muscle activation is of major interest in the study of human locomotion. From changes in the amplitude of surface electromyograms (EMGs) sampled with a pair of electrodes, indeed, our knowledge on the temporal profile of muscle excitation during gait and running has broadened [1][2]. Implicit in this logic is the premise that a single, bipolar EMG is a proxy of muscle excitation [3]. However, evidence garnered from the sampling of surface EMGs with multiple electrodes from a target muscle, the high-density EMG (HD-EMG), undermines the putative notion that inferences on the excitation of the whole muscle can be drawn from EMGs detected with a single pair of surface electrodes [4][5]. Only with HD-EMG can the Type I (low specificity–i.e., crosstalk) and the Type II (low sensitivity) errors in surface electromyography be minimized [6].

If EMGs are to be collected during locomotion, two issues must be contended with. First, in dynamic contractions, many are the confounding factors defying the use of EMG amplitude as a surrogate of muscle excitation [7]. For example, changes in muscle architecture as well as artifacts due to muscle and cable movements are expected to concur to changes in EMG amplitude [8] – [10]. Second, subjects must be able to move with minimal constraints while wearing electrodes. Otherwise, the detection system would itself constitute a confounding factor. In this regard, the systems commercially available for the acquisition of HD-EMG are specious, with encumbrance augmenting with the number of EMGs to be sampled per muscle. This second issue is aggravated when on-field studies of locomotor activities are required [11], [12] given subjects would be requested to wear the whole, HD-EMG acquisition system.

In this single-case study, we provide evidence that both issues may be overcome. By combining two systems we developed for the detection of high-resolution kinematics data [13] and HD-EMG [14] we specifically show that EMGs of high quality can be recorded from multiple locations of the gastrocnemius muscle during high-speed locomotion–80 m sprints. The quality of EMG was assessed by comparing features of the HD-EMG we obtained (e.g. propagation and representation of single action potentials; [6]) with those expected from our current knowledge in the field. Our results are expected to stimulate the emergence of on-field studies in extreme, dynamic conditions.

II. METHODS

A. Subject

A single, healthy subject volunteered to participate in this study (male, 24 years, 73 kg, 180 cm), after providing written informed consent. The subject is an elite runner, scoring over 700 points in the tables of World Athletics [15], which correspond to taking less than 11.66 s to cover the 100 m distance. No musculoskeletal injuries affecting his sprinting ability have been reported in the preceding six months. The experimental protocol was applied in agreement with the Declaration of Helsinki and were approved by the Regional Ethics Committee.

XXX-X-XXXX-XXXX-X/XX/$XX.00 ©20XX IEEE

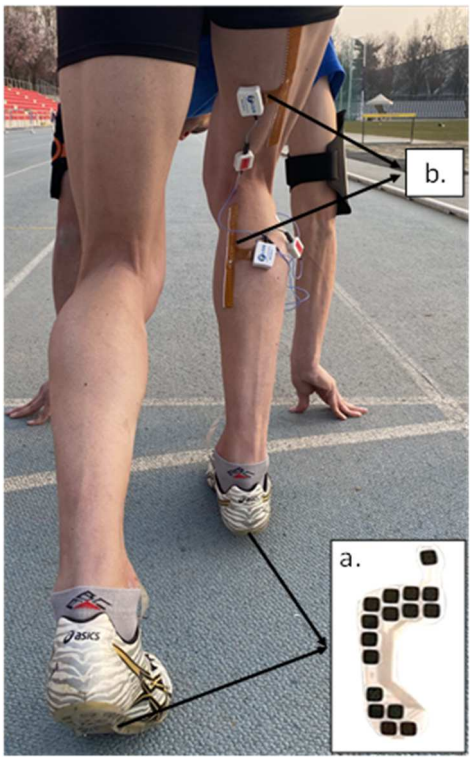

Fig. 1 Subject at the starting position, showing the systems used for the acquisition of kinematics data (a; sensoried insole) and of HD-EMGs (b).

B. *Runing protocol*

After warming up for 40 min to avoid muscle injury, the subject was asked to run at maximal speed three times. Intervals of at least 3 min between trials were applied. During the three sprint trials, starting from the crouch three-point position, the subject run for 80 m along the lane of an official, 400 m track. The time taken to cover the 80 m distance was measured with a hand chronometer (Sasso Marconi (Bo), Italia, Motus chronometry millennium MT50, sensitivity 1/1000). In the first sprint the subject did not wear anything other than the garment he typically wears during training. In the second and third sprints the subject wore the systems for the acquisition of kinematics data and of HD-EMG (section II.C). Under-sampled synchronization (488 µs sampling interval) was achieved using the system developed in [16].

C. *Acquisition of kinematics data and HD-EMG*

Kinematics data were acquired using the INDIP multi-sensor system [17]. FRS pressure insoles were placed in each of the running shoes selected by the subject for the running trials (mod. YETI, 221e S.r.l., Padua, Italy. The insole includes sixteen pressure sensors; 16 (element area = 310 mm^2; force threshold = 5 N) are distributed over each insole, with one sensor beneath the hallux and the remaining 15 sensors spanning the rearfoot and forefoot. Data from the insoles were collected concurrently with data from a three-axial magneto-inertial measurement unit [13], [18]. The control unit of each insole managed both the sampling (100 Hz) of pressure and inertial data and storage into a datalogger, fig. 1a illustrates the insoles.

The system recently developed in our laboratory [14] was used for the recording of HD-EMGs. Among the key features of this system, making it mostly suitable for on-field EMG acquisition in dynamic conditions, are the:

- Wireless communication with acquisition devices, from desktop computers to smart devices. The latter option would appear a mandatory feature for on-field, dynamic applications, wherein wearing the acquisition device is needed.
- Modular architecture, with each module sampling 32 EMGs (conditioned inputs) plus one unconditioned auxiliary input used for synchronization.
- Small size and weight of acquisition modules (34x30x15 mm, 17 g), facilitating the securing of modules to the skin.
- Absence of cables between the grid of electrodes and the modules, which helps assuaging movement artifacts due to the triboelectric effect between the conductors constituting the cables.

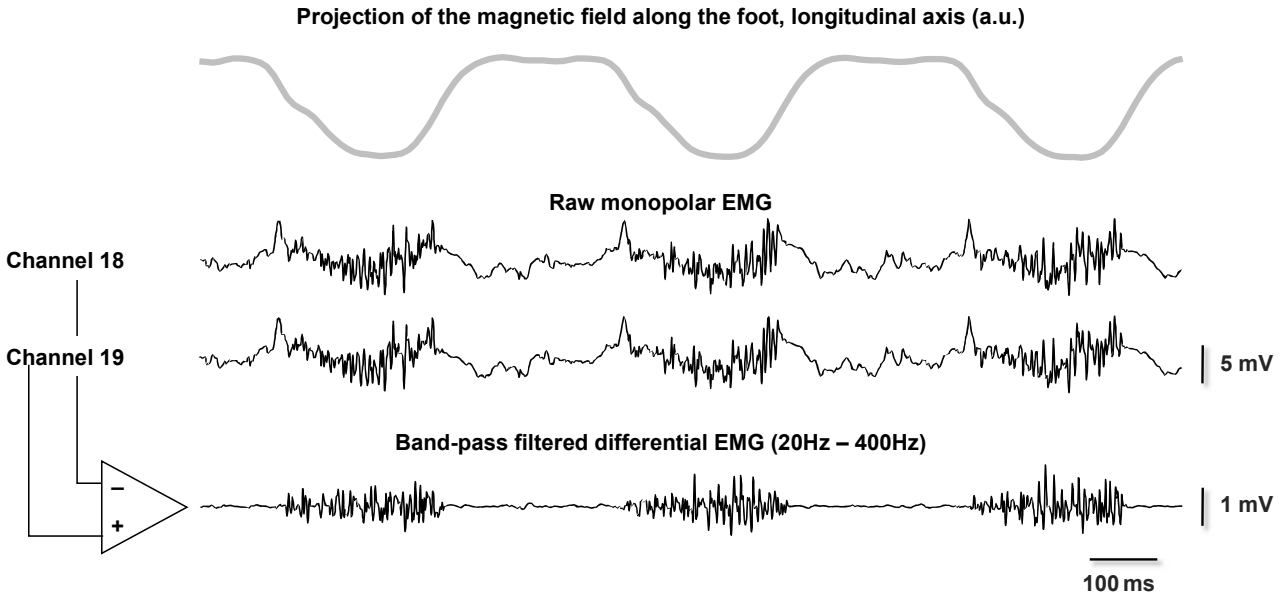

Fig. 2. Example of HD-EMG and of the magnetometer data along the foot longitudinal axis. Raw, monopolar EMGs from two consecutive electrodes are shown in the middle panel. The differential EMG resulting from the two monopolar signals is shown in the bottom panel, after being band-pass filtered (20–400Hz)

Thirty-two monopolar EMGs from the gastrocnemius and biceps femoris muscles of the right side were amplified (192 V/V) and then sampled at 2048 Samples/s with a 16-bit A/D converter. Data were acquired with a cellphone (Samsung Galaxy S20) and elastic bands were used to secure it to the subject arm. With router, server, and clients being all wearable, the subject was able to run comfortably while biomechanical and electrophysiological data were collected with minimal, if any, losses.

Data from the insole and EMG systems were offline synchronized by a common, trigger pulse (500 ms, 0V–5V), issued to both systems via Bluetooth (insoles) and a wireless synchronization system (EMG; [16]).

D. Data processing

The running cycle is defined by two initial contacts of the same foot with the ground. The detection of the initial contacts from pressure sensors is based on the method proposed by Salis et al.[13] and adapted to running speeds. Briefly, the instants of rising edges of the pressure insoles signals (i.e. instants of activation of the sensing elements) are selected. Then, clusters of at least two spatially close sensing elements which activate with a maximum chronological distance of 0.1 s are identified. An initial contact is defined as the last rising instant of the first sequentially activated cluster.

After ensuring power line interference was negligible, single-differential EMGs were computed from the monopolar signals. Differential EMGs were then band-pass filtered using a zero-lag Butterworth filter (fourth order, 20–400 Hz; Fig. 2). Signal quality was assessed by visually inspecting EMGs with respect to the expected features in the EMG literature–absence of movement artifact and the presence of motor units action potentials. Finally, differential EMGs were segmented according to the onsets of foot contact and RMS values were computed over 2% percentiles of each running cycle and then averaged across cycles. Only EMGs detected from skin regions covering the superficial aponeurosis were retained for analysis (cf. section III and Fig. 3). We limited analysis to the gastrocnemius muscle, for which obtaining EMGs of high-quality is a sedulous process [19],[20].

III. Results and Discussion

In this single-case study, only a descriptive account is provided on the quality of HD-EMG collected during the technically challenging, 80 m sprint. With the aim of verifying how credible the possibility of assessing single motor units from surface EMGs may be, Enoka [21] systematically evaluated whether results provided by the decomposition of surface EMGs corroborate the knowledge on motor unit properties accumulated from studies conducted with intramuscular recordings. Similarly, here we rely on the knowledge matured in the field to assess the quality of the HD-EMGs collected. From a visual inspection of the HD-EMG, recorded, processed, and then segmented according to running cycles identified from the kinematics data, we specifically assess two features of major interest: the presence of movement artifacts and the representation of action potentials of individual motor units.

A. Movement artifacts

As expected, low-frequency fluctuations in the baseline of monopolar EMGs were observed during running. Given the acquisition module was connected directly to the grid of

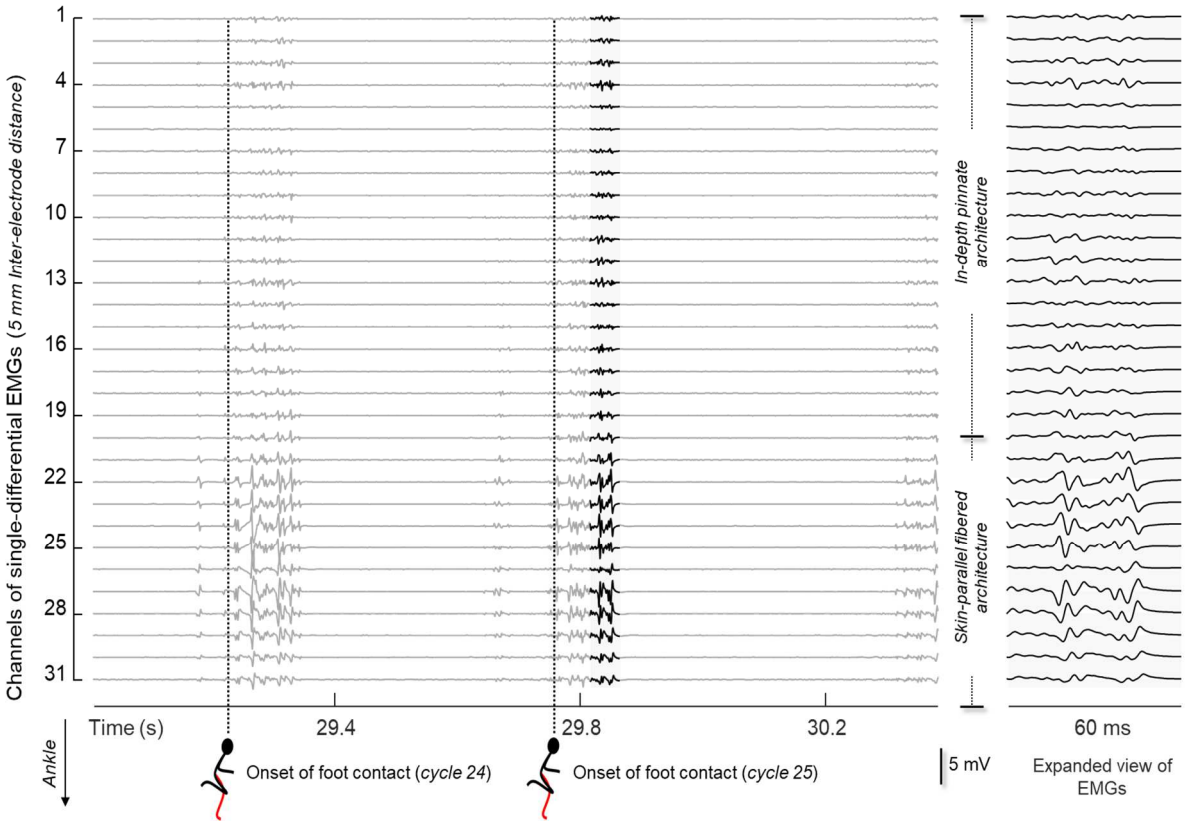

Fig. 3.HD-EMGs detected along the gastroncemius muscle during roughly two running cycles. Vertical, dashed lines indicate the onset of foot contact for two consecutive running cycles. An expanded view (60 ms) of the 31 differential EMGs is shown in the right, shaded panel.

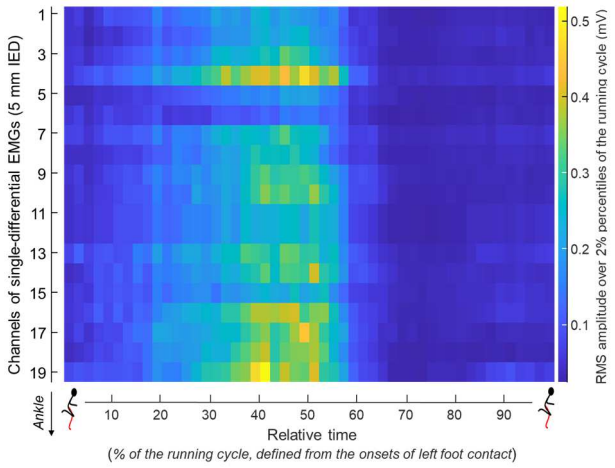

Fig. 4. Image created from the RMS value of EMGs colleced from all channels located over gastrocnemius superficial aponeurosis, over 2% of the running cycle.

electrodes (Fig. 1) and movement of the cable connecting the reference electrode to the acquisition module was suppressed, these baseline fluctuations are presumably due to relative movements between the muscle, the skin and the electrodes, leading to a fluctuation of the half-cell potential established at the electrode-skin interface. It is important to underline that this observation is exclusively related to the electrode-skin system and not to the acquisition system. Presumptive evidence of this assertion is the similar periodicity observed in the monopolar EMGs and in the projection of the Earth magnetic field along the foot longitudinal axis, as revealed by the inertial module (Fig. 1) placed dorsally on the forefoot: temporal changes in the magnetic field projection match closely the baseline fluctuations in the raw, monopolar signals (Fig. 2). It would appear tempting to speculate that this observation stimulates the emergence of technical studies aiming to explore the possibility of segmenting running cycles from EMGs.

While indicative of ankle kinematics during running, the fluctuations in EMG baseline overtly bias estimations of the degree of muscle excitation from EMG amplitude. Descriptors of EMG amplitude would be affected by the energy of this low-frequency component, leading to fairly high overestimation of the degree of muscle excitation. Nevertheless, the bandwidth of the movement artifacts we observed seemingly overlaps to marginal extents with that expected for the muscle fiber action potentials [22]. Indeed, the 20 Hz high-pass filtering recently recommended for the attenuation of movement artifacts [23],[24] seems to have suppressed movement artifacts with minimal attenuation of the physiological content in the differential EMGs (Fig. 2).

B. *Motor unit action potentials*

From an EMG perspective, gastrocnemius is a unique muscle. The arrangement of gastrocnemius fibers in relation to the skin changes along the muscle: between the distal extremities of the superficial and deep aponeuroses skin and fibers reside in parallel planes whereas, from the proximal to the distal extremity of the superficial aponeurosis, the fibers are disposed in an oblique plane relative to the skin [19]. We named these two peculiar arrangements between skin and fibers respectively as *skin-parallel fibered* and *in-depth pinnate* architectures [6].

The relevance of this within-gastrocnemius differential architecture lies in the information EMGs detected from both regions convey. Features typically observed in HD-EMGs detected from fusiform muscles (e.g. biceps brachii; [25]) are expected to be observed in HD-EMGs detected from the *skin-parallel fibered* region in gastrocnemius. Indeed, inspection of the EMGs we collected during running from the distal gastrocnemius region, where the *skin-parallel fibered* architecture manifests [19],clearly show (Fig. 3):

- Innervation zone in channel 26, characterized by an EMG with remarkably low amplitude and surrounded by EMG with opposed phase.
- Propagating potentials, as evinced by the delay between peaks in EMGs detected consecutively on either side, top or bottom, from the innervation zone.

Conversely, as anticipated, neither innervation zone nor propagating potentials were observed in EMGs detected from the *in-depth pinnate* fibers. In this region, surface EMGs detected by different electrodes represent the contribution of different fibers, occupying different locations along the muscle longitudinal axis [20]. As a consequence, the amplitude of surface EMGs detected from different points over the gastrocnemius superficial aponeurosis is associated with the number and location of the excited, *in-depth pinnate* fibers. While conduction velocity and innervation zone may be assessed distally from gastrocnemius, regional changes in gastrocnemius excitation can be assessed mid-proximally. Regardless of whether inspecting EMGs distally or proximally, during on-field sprints we observed surface action potentials with duration and shape similar to those reported for isometric, well-controlled contractions [20].

Once the quality of HD-EMGs has been ascertained, considerations of applied value may be made. Fig. 4 illustrates the spatio-temporal representation of the amplitude of HD-EMGs detected from the *in-depth pinnate*, gastrocnemius region (c.f. Fig. 3). For the elite athlete we tested, gastrocnemius was mostly excited during the second quarter of the running cycle, when EMG amplitude was generally greatest along the grid (Fig. 4). Slight, regional differences in amplitude were observed however, with greatest RMS values being represented at the muscle proximal and distal thirds. If changes in gastrocnemius architecture with ankle movement manifested equally along the muscle, the proximo-distal differences observed in EMG amplitude would possibly suggest the gastrocnemius was not maximally excited, notwithstanding the subject had run at his maximal speed. Otherwise, EMGs with more similar amplitude would have been expected to be detected across the channels. While inferences cannot be drawn from a single subject, our results encourage the use of HD-EMG to further our understanding on muscle function during highly dynamic contractions and outside the laboratory setting.

IV. Conclusions

In this study we investigated the possibility of recording high-quality EMGs with multiple electrodes during extreme, dynamic contractions–80 m sprints. Data were collected on an athletic track, with ad-hoc systems developed for on-field applications. Despite having tested a single subject, our results corroborate the temporal features expected for high-quality recordings in the EMG literature. Absence of movement artifacts (Fig. 2) and the presence of action potentials of single motor units throughout the running cycles

(Fig. 3) advocate the use of HD-EMG to study muscle function in highly dynamic contractions and in more ecological scenarios.

ACKNOWLEDGMENT

The work was supported by the MOBILISE-D (EU H2020, EFPIA, and IMI 2 Joint Undertaking; Grant no. 820820). The study sponsors were not involved in the study phases, in the writing of the manuscript and in the decision about its submission.

REFERENCES

[1] S. Jönhagen, M. O. Ericson, G. Németh, and E. Eriksson, "Amplitude and timing of electromyographic activity during sprinting," Scandinavian Journal of Medicine and Science in Sports, vol. 6, no. 1, pp. 15–21, 1996, doi: 10.1111/j.1600-0838.1996.tb00064.x.

[2] H. Kyröläinen, J. Avela, and P. v. Komi, "Changes in muscle activity with increasing running speed," Journal of Sports Sciences, vol. 23, no. 10, pp. 1101–1109, Oct. 2005, doi: 10.1080/02640410400021575.

[3] A. D. Vigotsky, I. Halperin, G. S. Trajano, and T. M. Vieira, "Longing for a Longitudinal Proxy_ Acutely Measured Surface EMG Amplitude is not a Validated Predictor of Muscle Hypertrophy," Sports Medicine, vol. 52, no. 2, pp. 193–199, Feb. 2022, doi: 10.1007/s40279-021-01619-2.

[4] K. Watanabe, M. Kouzaki, and T. Moritani, "Regional neuromuscular regulation within human rectus femoris muscle during gait," Journal of Biomechanics, vol. 47, no. 14, pp. 3502–3508, 2014, doi: 10.1016/j.jbiomech.2014.09.001.

[5] N. G. G. Readi, V. Rosso, A. Rainoldi, and T. M. M. M. M. Vieira, "Do sweep rowers symmetrically activate their low back muscles during indoor rowing?," Scandinavian journal of medicine & science in sports, vol. 25, no. 4, pp. e339--52, Aug. 2015, doi: 10.1111/sms.12319.

[6] T. M. Vieira and A. Botter, "The Accurate Assessment of Muscle Excitation Requires the Detection of Multiple Surface Electromyograms," Exercise and Sport Sciences Reviews, vol. 49, no. 1, pp. 23–34, Jan. 2021, doi: 10.1249/JES.0000000000000240.

[7] D. Farina, "Interpretation of the Surface Electromyogram in Dynamic Contractions," 2006. [Online]. Available: www.acsm-essr.org

[8] J. R. Potvin, "Effects of muscle kinematics on surface EMG amplitudeand frequency during fatiguing dynamic contractions," Journal of applied physiology (Bethesda, Md. : 1985) , vol. 82, no. 1, pp. 144–51, 1997.

[9] T. M. Vieira, M. C. Bisi, R. Stagni, and A. Botter, "Changes in tibialis anterior architecture affect the amplitude of surface electromyograms," Journal of NeuroEngineering and Rehabilitation, vol. 14, no. 1, Aug. 2017, doi: 10.1186/s12984-017-0291-5.

[10] F. D. Mancebo, H. v. Cabral, L. M. L. de Souza, L. F. de Oliveira, and T. M. Vieira, "Innervation zone locations distribute medially within the pectoralis major muscle during bench press exercise," Journal of Electromyography and Kinesiology, vol. 46, pp. 8–13, Jun. 2019, doi: 10.1016/j.jelekin.2019.03.002.

[11] A. Mero and P. v. Komi, "Electromyographic activity in sprinting at speeds ranging from sub-maximal to supra-maximal," Medicine and Science in Sports and Exercise, vol. 19, no. 3, pp. 266–274, 1987.

[12] T. W. Dorn, A. G. Schache, and M. G. Pandy, "Erratum: Muscular strategy shift in human running: Dependence of running speed on hip and ankle muscle performance (Journal of Experimental Biology 215 (1944-1956))," Journal of Experimental Biology, vol. 215, no. 13. p. 2347, Jul. 01, 2012. doi: 10.1242/jeb.075051.

[13] F. Salis, S. Bertuletti, T. Bonci, U. della Croce, C. Mazzà, and A. Cereatti, "A method for gait events detection based on low spatial resolution pressure insoles data," Journal of Biomechanics, vol. 127, Oct. 2021, doi: 10.1016/j.jbiomech.2021.110687.

[14] G. L. Cerone, A. Botter, and M. Gazzoni, "A modular, smart, and wearable system for high density sEMG detection," IEEE Transactions on Biomedical Engineering, vol. 66, no. 22, pp. 3371–3380, Dec. 2019, doi: 10.1109/TBME.2019.2904398.

[15] B. Spiriev, "IAAF SCORING TABLES OF ATHLETICS," 2017.

[16] G. L. Cerone, A. Giangrande, M. Ghislieri, M. Gazzoni, H. Piitulainen, and A. Botter, "Design and Validation of a Wireless Body Sensor Network for Integrated EEG and HD-sEMG Acquisitions," IEEE Transactions on Neural Systems and Rehabilitation Engineering, vol. 30, pp. 61–71, 2022, doi: 10.1109/TNSRE.2022.3140220.

[17] F. Salis et al., "A wearable multi-sensor system for real world gait analysis," 2021 43rd Annual International Conference of the IEEE Engineering in Medicine & Biology Society (EMBC), pp. 7020–7023, 2021.

[18] R. Rossanigo, M. Caruso, F. Salis, S. Bertuletti, U. della Croce, and A. Cereatti, "An optimal procedure for stride length estimation using foot-mounted magneto-inertial measurement units," Jun. 2021. doi: 10.1109/MeMeA52024.2021.9478604.

[19] E. F. E. F. Hodson-Tole, I. D. I. D. Loram, and T. M. T. M. M. Vieira, "Myoelectric activity along human gastrocnemius medialis: Different spatial distributions of postural and electrically elicited surface potentials," Journal of Electromyography and Kinesiology, vol. 23, no. 1, pp. 43–50, 2013, doi: 10.1016/j.jelekin.2012.08.003.

[20] T. M. M. Vieira, I. D. Loram, S. Muceli, R. Merletti, and D. Farina, "Postural activation of the human medial gastrocnemius muscle: are the muscle units spatially localised?," Journal of Physiology-London, vol. 589, no. 2, pp. 431–443, Jan. 2011, doi: 10.1113/jphysiol.2010.201806.

[21] R. M. Enoka, "Physiological validation of the decomposition of surface EMG signals," Journal of Electromyography and Kinesiology, vol. 46, no. March, pp. 70–83, 2019, doi: 10.1016/j.jelekin.2019.03.010.

[22] A. Van Boxtel and L. R. B. B. Schomaker, "Motor Unit Firing Rate During Static Contraction Indicated by the Surface EMG Power Spectrum," IEEE Transactions on Biomedical Engineering, vol. BME-30, no. 9, pp. 601–609, 1983, doi: 10.1109/TBME.1983.325057.

[23] R. Merletti and G. L. Cerone, "Tutorial. Surface EMG detection, conditioning and pre-processing: Best practices," Journal of Electromyography and Kinesiology, vol. 54, 2020, doi: 10.1016/j.jelekin.2020.102440.

[24] B. R. Schlink, A. D. Nordin, and D. P. Ferris, "Comparison of signal processing methods for reducing motion artifacts in high-density electromyography during human locomotion," IEEE Open Journal of Engineering in Medicine and Biology, vol. 1, pp. 156–185, 2020, doi: 10.1109/OJEMB.2020.2999782.

[25] R. Merletti, D. Farina, and M. Gazzoni, "The linear electrode array: A useful tool with many applications," Journal of Electromyography and Kinesiology, vol. 13, no. 1, pp. 37–47, 2003, doi: 10.1016/S1050-6411(02)00082-2.