The Sensitivity of Bipolar Electromyograms to Muscle Excitation Scales With the Inter-Electrode Distance

Taian M. Vieira, Giacinto Luigi Cerone, *Member, IEEE*, Alberto Botter, *Member, IEEE*, and Kohei Watanabe, and Andrew D. Vigotsky

***Abstract*— The value of surface electromyograms (EMGs) lies in their potential to non-invasively probe the neuromuscular system. Whether muscle excitation may be accurately inferred from bipolar EMGs depends on how much the detected signal is both sensitive and specific to the excitation of the target muscle. While both are known to be a function of the inter-electrode distance (IED), specificity has been of long concern in the physiological literature. In contrast, sensitivity, at best, has been implicitly assumed. Here we provide evidence that the IED imposes a biophysical constraint on the sensitivity of surface EMG. From 20 healthy subjects, we tested the hypothesis that excessively reducing the IED limits EMGs' physiological content. We detected bipolar EMGs with IEDs varying from 5 mm to 50 mm from two skeletal muscles with distinct architectures, gastrocnemius and biceps brachii. Non-parametric statistics and Bayesian hierarchical modelling were used to evaluate the dependence of the onset of muscle excitation and signal-to-noise ratio (SNR) on the IED. Experimental results revealed that IED critically affects the sensitivity of bipolar EMGs for both muscles—indeliberately reducing the IED yields EMGs that are not representative of the whole muscle, hampering validity. Simulation results substantiate the generalization of experimental results to small and large electrodes. Based on current and previous findings, we discuss a potentially valid procedure for defining the most appropriate IED for a single bipolar, surface recording—i.e., the distance from the electrode to the target muscle boundary may heuristically serve as a lower bound when choosing an IED.**

***Index Terms*— Excitation onset, muscle excitation, surface electrodes, transfer function.**

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This work involved human subjects or animals in its research. Approval of all ethical and experimental procedures and protocols was granted by the Regional Ethics Committee (Commissione di Vigilanza, Servizio Sanitario Nazionale-Regione Piemonte-ASL 1-Torino, Italy), and performed in line with the Declaration of Helsinki.

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I. Introduction

SURFACE electromyography (EMG) is a powerful technique to non-invasively assess the neuromuscular system. To do so, electrodes are placed on top a muscle, and electrical potentials are measured in monopolar or bipolar derivation. The configuration of these electrodes will ultimately affect the sampled signal. As such, different systems of electrodes have been used for sampling surface EMG, varying in terms of size, number and distribution of electrodes [1], [2], [3], [4], [5]. In particular, grids of electrodes offer the possibility of mapping the spatial distribution of EMG amplitudes, in turn enabling the assessment of regional excitation within the muscle [2], [6], [7], [8] when used properly [9], [10]. Despite their methodological and inferential value, electrode grids are not commonplace in surface EMG studies; instead, single bipolar setups dominate.

Bipolar electrodes are ubiquitous in the surface EMG literature. Relative to grid electrodes, they are easier to use, affordable, and regularly available. Based on the amplitude of EMGs detected using bipolar configurations, information on the timing and degree of muscle excitation can be obtained [9]. Yet, how accurately the amplitude of bipolar EMGs reflects the neurophysiological events underpinning muscle excitation depends on the extent to which the detected signal is confounded [11]. There are two principal effects of confounding.

Specificity and sensitivity are two competing issues that determine the validity of inferences that can be drawn from bipolar EMGs. Ideally, the amplitude of bipolar EMGs reflects the net excitation of the entire target muscle and nothing else. On occasion, electrodes may sample from other, nearby muscles, meaning that the EMG is not specific to the target muscle. This could inevitably lead to inferring muscles are excited when they are not (Type I error; [9]). This lack of specificity, often referred to as crosstalk, has long been a major concern in electromyography [4], [12], [13]. On the other hand, electrodes may sample from a small, unrepresentative fraction of the

target muscle, introducing a bias in the sensitivity of EMGs to changes in muscle excitation. In other words, inferring that the muscle is not excited when it actually is would be the logical consequence of collecting surface EMGs that lack sensitivity (Type II error: [9]). The contexts in which these errors occur and are problematic depend on both the research question and experimental parameters.

There is an inherent tradeoff between sensitivity and specificity. Which of the two issues predominates in bipolar recordings depends on the amount of biological tissue sampled by the surface electrodes and, thus, on the inter-electrode distance (IED). Short IEDs are more likely to provide more specific and less sensitive EMGs, whereas the opposite holds true for large IEDs [14], [15], [16]. While specificity is generally acknowledged in the EMG literature, with recommendations tending to favor the use of short IEDs [13], only recently has the sensitivity issue been systematically studied [3], [14]. Although the existence of Type II errors in surface EMG has been formalized [9], its practical importance has not yet been documented. More specifically, how much does sensitivity affect the inferences we draw regarding muscle excitation when using bipolar surface EMGs?

In this study, we use a dense array of surface electrodes to systematically assess this issue for two different muscle geometries. With this array, we were able to compute bipolar EMGs for progressively greater IEDs, from 5 mm to 50 mm, with all pairs of electrodes being centered roughly at the same skin region over biceps brachii (BB) and gastrocnemius medialis (GM). From these signals, we specifically investigate how much the onset of muscle excitation and the quality of surface EMGs (signal-to-noise ratio; SNR) are affected by IED. If EMG sensitivity is a concern for the estimation of muscle excitation, we would expect EMGs detected by shorter IEDs to provide significantly delayed onsets, highly variable onsets between subjects or both (cf. Fig 2 in [17]). We also expect the SNR values to increase with IED, possibly saturating for IEDs after which bipolar EMGs become a scaled version of monopolar EMGs [18]. From EMGs simulated with validated models [19], [20], we assessed the validity of experimental results for electrodes of different sizes. Based on previous, theoretical accounts [19], [21], we anticipate the tradeoff between sensitivity and IED holds regardless of the bipolar electrodes' size. Importantly, we do not intend for this manuscript to downplay the relevance of crosstalk. Instead, for the first time, we wish to document the effect of low sensitivity in surface recordings. In doing so, we provide direct evidence for the sensitivity-specificity tradeoff that is a function of IED.

II. Methods

A. Participants

Twenty healthy subjects (7 women) volunteered to participate in this study (range values; age: 21-38 years; body mass: 48-88 kg; height: 158-187 cm). All subjects provided written informed consent prior to inclusion in the study and after being informed of the experimental procedures. The experimental protocol conformed with the Declaration of Helsinki and was approved by the Regional Ethics Committee (Commissione di Vigilanza, Servizio Sanitario Nazionale-Regione Piemonte-ASL 1-Torino, Italy).

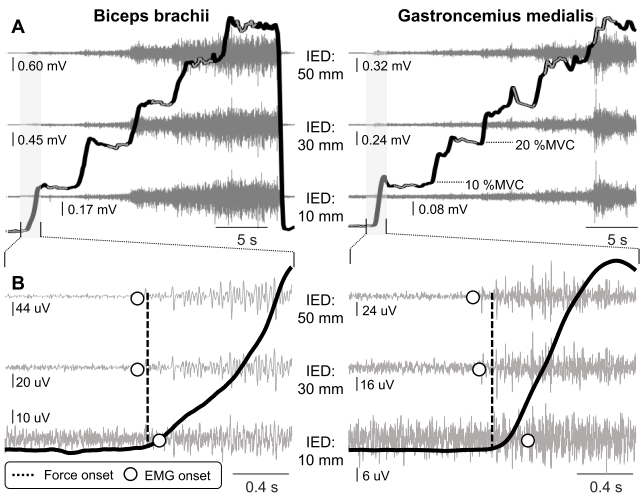

Fig. 1. Raw surface EMGs (grey traces) and torque data (black traces) are shown for the staircase, isometric contractions, separately for biceps brachii (BB) and Gastrocnemius medialis (GM) muscles. Data are shown for almost the whole trial in panel **A**, with bipolar EMGs obtained for 10 mm, 30 mm and 50 mm inter-electrode distances (IEDs) shown respectively from bottom to top. Note the different scales used to represent EMGs for different IEDs. An expanded portion of signals centered roughly at the torque onset (vertical, dashed line) is shown in **B**. EMG onsets computed for each IED (cf. Methods) are represented with white circles.

B. Experimental Protocol

Two experimental sessions were conducted. In the first session, subjects were asked to sit comfortably with their forearm lying over a brace built to measure elbow flexion moments [22]. The elbow rotation axis was aligned coaxially with the axis of rotation of the torque meter, with the shoulder abducted at 45° and the elbow flexed at 90° with the wrist supine. After securing the wrist to the torque device, three maximal voluntary contractions (MVCs) were applied, lasting 5 s each and with 2 min rest in-between. The highest torque value was then used to scale a staircase torque profile (Fig. 1A), comprising seven constant-force contractions. Specifically, the staircase consisted of 5s isometric contractions at 0%, 10%, 20%, 30%, 40%, 50% and 0% MVC [4], for a total of 35s of acquisition. Acquisition started only after the experimenter ascertained the validity of 0% MVC, defined as the absence of motor unit action potentials in the real-time display of raw monopolar surface EMGs. Ensuring participants were at rest upon the acquisition start was necessary for defining the EMG baseline (cf. section II-E). Three staircase force-profile trials were applied, with breaks of 5 min.

The same procedure was applied in the second experimental session, in which we measured ankle plantar flexion rather than elbow flexion moments. Subjects were seated comfortably, with the knee fully extended, the ankle at neutral position and the foot secured firmly to a torque device specifically designed for measuring ankle moments (Neg1, OTBioelettronica, Torino, Italy). The two sessions were applied in this fixed order, with roughly 30 min being required

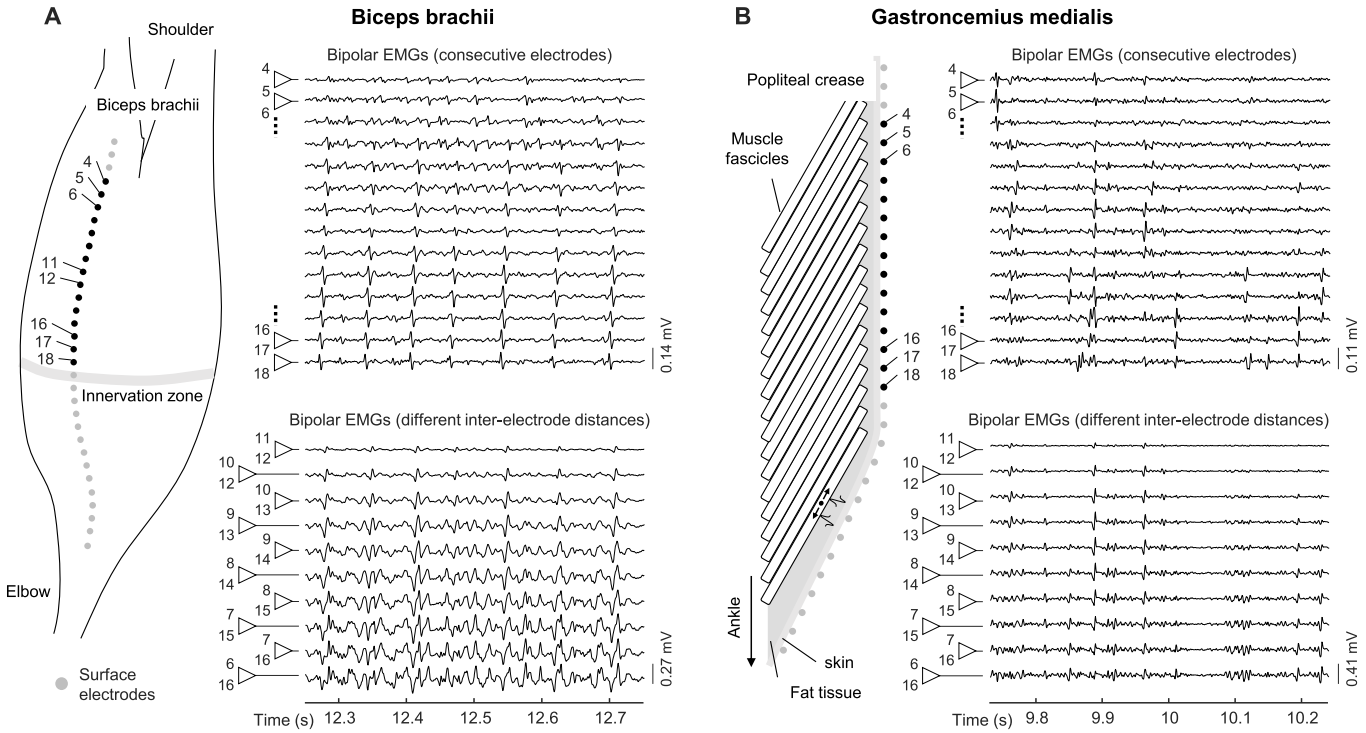

Fig. 2. Schematic representation of the procedure considered for positioning electrodes and obtaining bipolar EMGs for different IEDs, shown separately for BB (**A**) and GM (**B**). For BB, only electrodes located proximally to the innervation were considered for analysis (black circles) whereas, for GM, only electrodes located over the muscle superficial aponeurosis were considered. For both muscles, bipolar EMGs were created from monopolar signals detected by consecutive electrodes (top panels) and by electrodes located at progressively greater distances between themselves, from 5 to 50 mm IEDs (bottom panels; (1)).

to prepare the subject and to position electrodes between sessions.

C. EMG and Torque Measurements

Monopolar surface EMGs were collected with a flexible array of 32 circular electrodes ([14]; 5 mm inter-electrode distance; 2 mm diameter). Bi-adhesive pads were used to secure the electrode array to the skin after cleaning the skin with abrasive paste. Conductive paste was deposited in the cavities of the adhesive pads to ensure the electrical contact between the skin and the electrodes. Signals were amplified by a variable factor, ranging from 1,000 to 5,000, to provide the highest SNR without saturation (10-750 Hz bandwidth, EMG-USB amplifier, OTBioelettronica, Torino, Italy). Then, EMGs were digitized at 2048 Hz with a 12-bit A/D converter and stored for analysis. Torque signals were amplified (Forza, OTBioelettronica, Torino, Italy) and sampled synchronously with EMGs (EMG-USB amplifier).

The electrode array was positioned at specific locations for each muscle. We used ultrasound (Echo Blaster, Telemed UAB, Lithuania) to help identify the boundaries of the long head of BB ([23]; session 1) and of GM ([6]; session 2), which after being identified were marked on the skin. Electrodes were aligned parallel to the muscle longitudinal axis, with the center of the array being positioned halfway between the proximal and distal myotendinous junctions and halfway between the most medial and lateral aspects of both muscles. A schematic representation of the position of electrodes relative to each muscle is presented in Fig. 2.

D. Computing Bipolar EMGs for Different IEDs

Single-differential (bipolar) EMGs were computed from the monopolar signals [9] detected by ten different pairs of electrodes, centered roughly at same location along the muscles and with progressively larger IEDs (Fig. 2). For the BB muscle, the first pair of electrodes (e_i, e_{i+1}) selected was that located halfway the muscle innervation zone, identified as described by [24], and the proximal myotendinous junction, identified with ultrasound imaging ($e_{i=11}$ and $e_{i+1=12}$ in Fig. 2A). At this point, electrodes located progressively more proximal and distal to the initial pair of electrodes were used to compute the bipolar EMGs (BP^k_{emg}):

$$BP^k_{\text{emg}}[\text{n}] = V_{\text{e}_{i+1+d[k]}}[\text{n}] - V_{\text{e}_{i-\text{p}[k]}}[\text{n}] \quad (1)$$

where V_{e} corresponds to the voltage reading for band-pass filtered (4^{th} order Butterworth, 15-350Hz) monopolar EMGs and n indicates the sample number, ranging from 1 to 71680 (35s of acquisition). Different IEDs are represented by $p = 0, 1, 1, 2, 2, 3, 3, 4, 4, 5$ and $d = 0, 0, 1, 1, 2, 2, 3, 3, 4, 4$, respectively denoting the number of electrodes to shift proximally and distally for computing the bipolar EMGs with progressively greater IEDs. With this approach, IED varied from 5 to 50 mm at steps of 5 mm ($IED[k] = 5(1 + d[k] - p[k]), k \in \{1, \ldots, 10\}$). The same procedure was applied to GM, although for this muscle the first pair of electrodes considered was that centered halfway between the distal extremity of the superficial aponeurosis and the proximal myotendinous junction (Fig. 2B).

E. Assessing IED Effect on EMG Descriptors

We assessed the effect of IED on the timing and degree of muscle excitation, both inferred from the amplitude of EMGs obtained for the 10 different IEDs. The onset of muscle excitation was assessed for full-wave rectified and low-pass filtered (2nd order Butterworth filter; 30 Hz cut-off) bipolar EMGs: the EMG envelopes (BP^k_{env}). Briefly, we first calculated the baseline torque threshold during rest, defined as the mean plus three times the standard deviation of torque values over the first 3s of acquisition [25]. The sample (n_{TO}) after which the torque data first exceeded this threshold was considered to define the torque onset (Fig. 1B) and the search interval for the EMG onset [26]. Using the same procedure for the torque data, we computed the baseline EMG threshold ($BP^k_{\text{threshold}}$) from the EMG envelopes. Then, we computed the transition index (TI) by iteratively counting the number of samples not exceeding (N_{low}) and exceeding (N_{high}) $BP^k_{\text{threshold}}$ for each IED over a 200 ms window (N_w=410):

$$TI[j] = N^k_{\text{low}}[j] + N^k_{\text{high}}[j] \tag{2}$$

$$N^k_{\text{low}}[j] = \sum_{n=2}^{N_w+1} \text{Ind}\{BP^k_{\text{env}}[j-n+w] < BP^k_{\text{threshold}}\} \tag{3}$$

$$N^k_{\text{high}}[j] = \sum_{n=0}^{N_w-1} \text{Ind}\{BP^k_{\text{env}}[j+n+w] > BP^k_{\text{threshold}}\} \tag{4}$$

$$\text{Ind}(x) = \begin{cases} 1 & \text{if } x \text{ is true,} \\ 0 & \text{otherwise} \end{cases} \tag{5}$$

where $w = n_{\text{TO}}$-410 samples (200 ms) and $j = 1, \ldots, 2458$, providing TI values from 200 ms prior to 1000 ms after the torque onset. EMG onset was then defined as the time sample where TI peaked. Transitions are defined by shifts across a threshold; the TI assesses this by summing the number of data points less than the threshold before time j with the number of data points greater than the threshold following time j. Thus, the point at which TI is maximized is when the transition occurs. We computed the onset for such a long search space (1200 ms) to account for earliest possible occurrences of EMG onset and occurrences of highly delayed, onset values (cf. Fig. 1B). For each IED, we computed EMG onset only when the mean amplitude of EMG envelopes over 1s after n_{TO} was greater than $BP^k_{\text{threshold}}$. Subjects not meeting this criterion for a given IED were not eligible for onset estimation.

The IED effect on the potential quality of estimates of the degree of muscle excitation was assessed based on SNR values. First, we computed the root mean square (RMS) amplitude over the first 3s of acquisition for each IED ($\text{RMS}^k_{\text{noise}}$). Then, we computed SNR values for RMS amplitudes calculated over 3s windows centered at periods corresponding to each of the five contraction levels applied ($\text{RMS}^k_{\%\text{MVC}\in 10,20,30,40,50}$):

$$\text{SNR}^k_{\%\text{MVC}} = 20\log_{10}\left(\frac{\text{RMS}^k_{\%\text{MVC}}}{\text{RMS}^k_{\text{noise}}}\right) \tag{6}$$

We understand (6) provides an overestimation of the actual SNR values given that both noise and signal contribute to $\text{RMS}^k_{\%\text{MVC}}$. Our decision to compute SNR as indicated here was based on the absence of signal for some subjects ($\text{RMS}^k_{\%\text{MVC}} \approx \text{RMS}^k_{\text{noise}}$), in particular for the shortest IED, and on our ultimate goal which does not depend on perfect estimates of SNR. We do not expect the bias introduced by not accounting for the contribution of noise to $\text{RMS}^k_{\%\text{MVC}}$ to depend on IED.

F. Simulating Larger Electrodes

Assessing the relationship between EMG sensitivity and IED for bipolar electrodes of different sizes placed on the same skin region is experimentally unviable. Therefore, we used validated models to simulate EMGs detected with bipolar electrodes of different sizes and spacing from BB [19] and GM [20]. Detailed information on the simulation procedures is provided in the corresponding articles. Briefly, we simulated bipolar EMGs for different combinations of electrode diameters (1 mm, 5 mm, and 10 mm) and IEDs (from 2 mm to 30 mm at 1 mm steps). For a given IED, only electrodes with diameters ≤ IED were simulated.[1] For the BB muscle, EMGs were simulated for populations of fibers placed according to a Gaussian distribution with a 2.5 mm standard deviation centered at three different depths, 5 mm, 10 mm, and 20 mm, one at a time (Fig. 3A). For GM, the standard deviation of the Gaussian distribution describing the location of the fibers' superficial endings was 20 mm, with all pairs of simulated bipolar electrodes being centered at three different distances from the most proximal, fiber ending: 0 mm, 6 mm, and 15 mm, one at a time (Fig. 3B). The spread of fibers within the simulated muscle volume was chosen according to the local representation of muscle excitation and movement in surface EMG [6], [17] and in ultrasound imaging [27], respectively. Changes in the sensitivity of EMGs with electrode size and IED were assessed by inspecting the RMS amplitude of each simulated bipolar EMG.

G. Statistics

Contingency tables were used to assess how often onsets could be estimated for the different IEDs. The inter-quartile interval was computed to assess the variability of onset estimates across IEDs, with significance being tested with the Spearman correlation analysis.

The effect of IED and contraction level on the quality of EMGs (i.e., SNR) was assessed with a Bayesian nonlinear hierarchical model, using `CmdStanR` via `brms` with default priors [28], [29], [30]. The model took the following form:

$$\text{SNR}_{ijkl} = \theta_{ijk}(1 - \exp(-\text{IED}_{ijkl} \cdot \lambda_{ijk})) + \varepsilon_{ijkl} \tag{7}$$

$$\log(\theta_{ijk}) = \gamma^{\theta}_{0ijk} + \alpha^{\theta}_{1i}x^{\text{EF}}_{ijk} + \alpha^{\theta}_{2i}x^{\text{PF}}_{ijk} + [\alpha^{\theta}_{3i}x^{\text{EF}}_{ijk} + \alpha^{\theta}_{4i}x^{\text{PF}}_{ijk}]log(x^{\text{MVC}}_{ijk}/30) \tag{8}$$

$$\log(\lambda_{ijk}) = \gamma^{\lambda}_{0ijk} + \alpha^{\lambda}_{1i}x^{\text{EF}}_{ijk} + \alpha^{\lambda}_{2i}x^{\text{PF}}_{ijk} + [\alpha^{\lambda}_{3i}x^{\text{EF}}_{ijk} + \alpha^{\lambda}_{4i}x^{\text{PF}}_{ijk}]log(x^{\text{MVC}}_{ijk}/30) \tag{9}$$

$$\alpha_{[1-4]i} = \beta_{[1-4]} + \gamma_{[1-4]i} \tag{10}$$

[1] In simulation, the issue of short circuit does not apply. Bipolar EMGs computed for IEDs matching the electrode diameter can be regarded as if the difference between the simulated IED and electrode diameter was infinitesimally greater than zero.

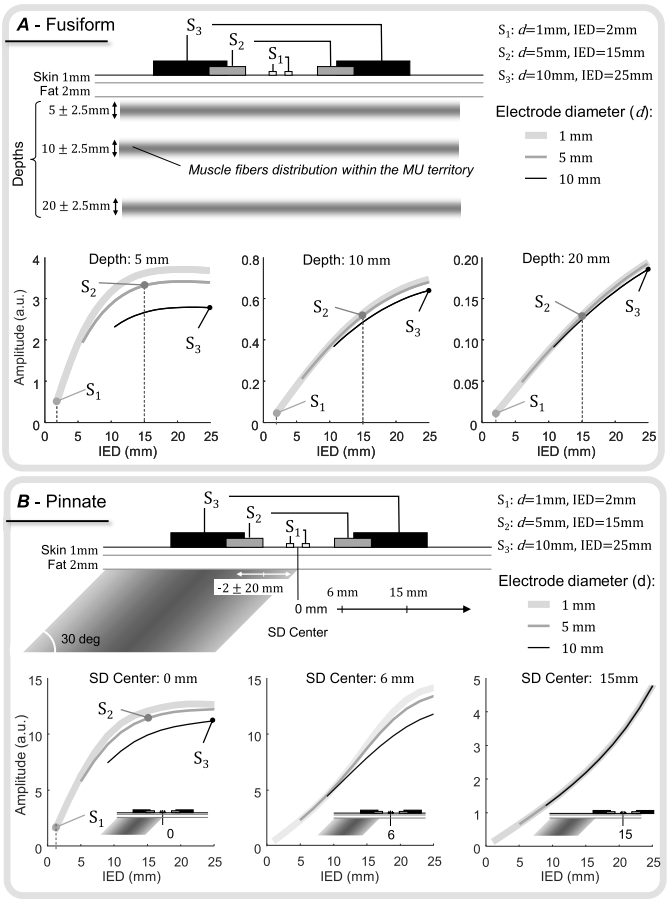

Fig. 3. The relationship between EMG amplitude and IED is shown for BB (A) and GM (B) muscles, considering bipolar, circular electrodes of three different diameters (d) represented with different line patterns (thick grey line: $d = 1$ *mm*; dark grey line: $d = 5$ *mm*; black line: $d = 10$ *mm*). Different distances between the center of the pair of electrodes and the population of excited fibers were simulated by considering different depths of fibers for BB (5, 10, and 20 mm) and by shifting electrodes proximally for GM (0, 6, 15 mm). For all simulated conditions, EMG amplitude increased with IED.

In the above, the subscripts i, j, and k are independent of how they were defined above in our data processing and reduction steps; this inconsistency across sections improves the readability and interpretability of this section. Here, the subscripts indicate the subject i, muscle group j (plantar flexors or elbow flexors), trial k, and IED measurement l. βs are population-level (or so-called "fixed effects") and γs are individual- and trial-level effects (or so-called "random effects"), depending on their subscripts. Equation 7 is Level 1 of the model, which models the SNRs within each trial as an exponential decay, starting at 0 and plateauing (asymptote) at θ with a decay or rate constant of λ. For convenience, we transformed the posterior samples of λ to the IED unit ($\tau^{\text{IED}}_{ijk} = \frac{1}{\lambda_{ijk}}$), which is analogous to a time constant in an RC circuit. For communication's sake, we call τ^{IED} the *distance constant*. Either λ or τ^{IED} could be used to determine how quickly SNR values converge to θ for increasing IEDs.

The model was fit using four chains of Hamiltonian Monte Carlo, each with 2,000 iterations: 1,000 for warmup and 1,000 for sampling with a thinning rate of 1, for a total of 4,000 samples per parameter. Model fits, convergence, and properties were evaluated using posterior predictive checks, trace plots, $\hat{R}$, and effective sample sizes. Notably, all parameters had $1 \leq \hat{R} \leq 1.01$. Since the parameters of interest were fit on the log scale, we exponentiated their posteriors to obtain their estimates on the more interpretable multiplicative scale. Effect estimates and their posterior distributions were interpreted continuously and probabilistically rather than dichotomizing whether effects differ from zero.

TABLE I
SECOND LEVEL POSTERIOR PARAMETER ESTIMATES FOR SNR VS. IED MODEL. POSTERIOR MEAN (95% CREDIBLE INTERVAL, CRI)

	SNR plateau (θ, dB)	SNR τ^{IED} (mm)
Biceps brachii		
Intercept	30 (27, 33)	5.0 (4.2, 5.8)
log(MVC/30)	1.50 (1.43, 1.57)	0.62 (0.54, 0.71)
Gastrocnemius medialis		
Intercept	25 (23, 26)	13.8 (10.2, 18.5)
log(MVC/30)	1.33 (1.25, 1.40)	0.58 (0.49, 0.68)

The parameters presented here are exponentiated to represent the multiplicative effects rather than the additive effects on the log scale. For example, biceps brachii has an expected plateau of 30 when MVC=30; however, if we increase MVC to 50, we get $15 \cdot 1.50^{log(50/30)} \approx 18$. *The same is true for GM and the distance constant model.*

III. RESULTS

Onset results from one subject in session 1 (N =19 subjects) were discarded because the amplitude of EMGs within 1s from the torque onset was below the baseline level for all IEDs. For one subject in session 2 (N =19 subjects) we were unable to compute onsets values because torque data was not collected, due to improper setting of the torque amplifier. For the 20 participants we correctly processed SNR.

A. IED vs. EMG onset: Experimental Results

The number of subjects eligible for onset estimates increased with IED (Fig. 4A). For both muscles, EMG onsets could be estimated for nearly all subjects for IEDs greater than 30 mm. For shorter IEDs, we were often unable to estimate onsets, in particular for the GM muscle, which provide 7 out of 19 valid estimates for 5mm IED. In addition to decreasing the number of eligible subjects, shorter IEDs frequently provided higher and highly variable onset estimates (Fig. 4A; cf. Fig. 1B for a representative subject). Interquartile intervals for EMG onsets were roughly three times smaller for IEDs greater than 30 mm when compared to 5 mm IED for both muscles (Fig. 4B; Spearman $Rho < -0.85$; $P < 0.005$).

B. IED vs. SNR values: Experimental Results

The quality (SNR) of EMGs was strongly affected by both IED and contraction level (Table I). For BB and GM, greater contraction levels and larger IEDs resulted in appreciably greater SNR values (Fig. 5A,B). Our hierarchical model was able to parse out the effects of muscle and contraction level; that is, the shape of the SNR-IED relationship differs

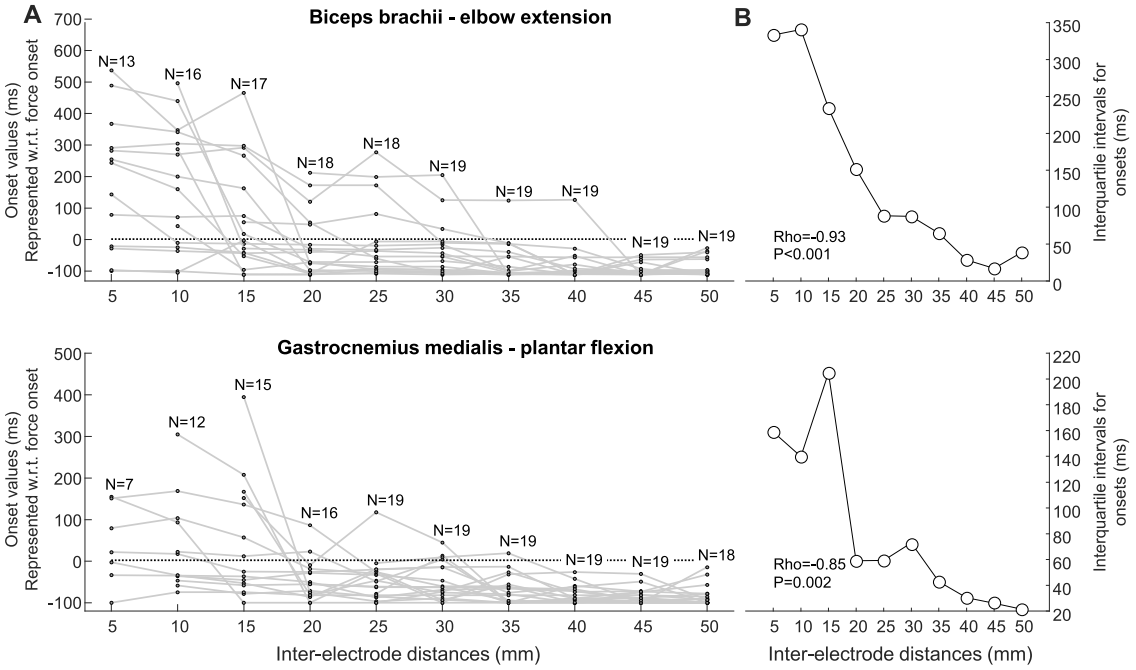

Fig. 4. Estimates of EMG onset for each subject (grey traces) and IED are shown in **A**, for BB (top) and GM (bottom). Numbers shown on top of each IED indicates the number of eligible subjects for onset estimation. For both muscles, 19 subjects were potentially eligible for onset estimation. Onset values were normalised in relation to the smallest onset value obtained for each subject across IEDs and are represented w.r.t. the torque onset averaged across subjects (horizontal, dashed line); EMG onsets below this value indicate increases in EMG amplitude precede increases in joint torque. The interquartile interval of onset values computed for each IED is shown in **B** (white circles), together with Spearman Rho and its P value.

depending on the muscle and contraction level. BB has greater plateaus than GM, especially at greater contraction levels (Fig. 5C). Moreover, SNR varies less steeply with IED for BB than GM, and this is more pronounced at lower MVCs (Fig. 5D). This results in BB having a greater rise SNR rates and greater SNR plateaus, resulting in greater SNRs for any given IED and MVC.

C. IED vs. Electrode Size: Simulation Results

For the two muscle architectures simulated—skin parallel fibered and in-depth pinnate [8]—IED affected sensitivity regardless of electrode size. How much sensitivity changed depended on the muscle and on the distance between electrodes and simulated fibers. For BB, the amplitude of bipolar EMGs decreased monotonically with IED shortening, for the four electrode sizes and for the three depths simulated (Fig. 3A). Similarly, the amplitude of EMGs simulated for GM decreased monotonically with IED shortening, regardless of the electrode diameter and distance to the excited muscle region (Fig. 3B). Even though the number of possible IEDs to simulate overtly decreases with electrode size, the monotonic decrease in EMG amplitude with IED shortening was evident even for the largest electrode simulated (10 mm diameter). Moreover, greater electrodes dramatically attenuated the EMG amplitude arising from the excitation of more superficial fibers for BB and from bipolar electrodes centered closer to the GM fibers (cf. offset between curves in Fig. 3). Most importantly, in spite of the attenuating effect of larger electrodes on the electric potential resulting from the excitation of more superficial fibers, the relationship between IED and EMG amplitude persisted across electrode sizes (Fig. 3).

IV. Discussion

In this study, we systematically assessed the effect of IED on excitation onset and on SNR when sampling surface EMGs with the traditional, bipolar montage. Using data from multiple monopolar EMGs collected along two muscles with different architectures, we computed bipolar EMGs for progressively greater IEDs, from 5 mm to 50 mm. Results from 20 subjects revealed the influence of IED on the onset of muscle excitation (Fig. 4) and the quality of the detected signals (Fig. 5). The greater the IED, the more sensitive the bipolar signal was to changes in joint torque-short IEDs resulted in EMG descriptors of dubious physiological validity. Our results suggest that indeliberately shortening IED is likely to inflate Type II error rates when using bipolar surface EMGs collected from the biceps brachii and the gastrocnemius muscles. From signals simulated with validated models, we observed the relationship between IED and EMG sensitivity to extend to bipolar electrodes of different sizes (Fig. 3). Based on our current and previous findings, we present and discuss a potentially generally valid procedure for defining the most appropriate IED in single bipolar, surface EMG studies.

A. Technical Considerations That Motivate Our Study and Justify Our Interpretation

Before interpreting our results, we wish to ensure readers are aware of the technical impetus of our study.

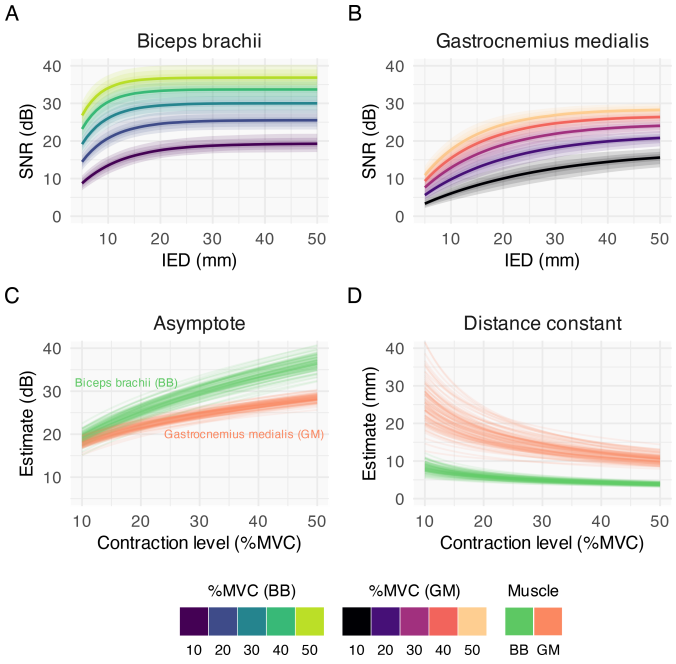

Fig. 5. Effects of IED, joint moment, and muscle on SNR. Plotted are expected posterior predictive distributions (ePPD) from our Bayesian nonlinear hierarchical model. (**A-B**) SNR increases with IED and joint moments MVC. (**C**) The asymptote of the SNR-IED relationship increases with MVC and differs between muscles. Although both muscles' asymptotes begin at 10% MVC, BB increases more steeply than GM. (**D**) Rates of exponential grow of the SNR-IED relationship are also a function of MVC and differ between muscles. BB has a higher growth rate (shorter distance constant) than GM, and both muscles' growth rates increase with increasing MVC. Shades of the error ribbons in (**A**) and (**B**) indicate the 50, 80, and 95% credible intervals of the ePPD. Each line in (**C**) and (**D**) is a draw from the ePPD.

Although consensus exists regarding the effect of IED on the EMG amplitude [4], [14], [15], [18], [31], precise figures for optimal and generally valid IEDs in single, bipolar recordings do not. IEDs $\leq$ 5 mm have been recommended for the specific case of studying action potential propagation using surface EMGs [32]. Indeed, this rather short IED threshold was not postulated to increase the representation of the target muscle excitation–that is, to increase sensitivity-or to suppress the representation of sources other than the target muscle-that is, to increase specificity-in the surface EMG. Rather, its goal was to attenuate the effect of non-propagating components on the estimation of muscle fiber conduction velocity [33]. In contrast, if researchers wish to extract the degree and timing of muscle excitation from a single, bipolar recording, such a short IED would appear unjustified. Therefore, for bipolar recordings, we propose the 'optimal' IED should simultaneously maximize (or balance) the sensitivity and specificity of EMGs' reading of excitation of the target muscle.

From a theoretical viewpoint, there are two salient technical issues that are necessary to understand to assess and thus choose an optimal IED. First, we should consider how IED affects the transfer function of the bipolar montage. From the theory of digital filters [34], the transfer function of a single pair of electrodes can be written as:[2]

$$H(z) = 1 - z^{-1} = 1 - e^{-2\pi f \mathrm{IED}} \tag{11}$$

with f corresponding to the spatial frequencies in the center-to-center direction of the two electrodes. Based on the Euler and the half-angle identities, and as reported by Reucher et al. [31], the magnitude of the electrodes' transfer function is defined by:

$$|H(z)| = 2\,\sin(\pi f \mathrm{IED}) \tag{12}$$

Since $|H(z)|$ increases monotonically from $f = 0$ cy/mm to $f = \frac{1}{2\,\mathrm{IED}}$cy/mm, bipolar electrodes will high-pass filter the electric potential distributed over the skin's surface. Most importantly, (12) establishes a dependence of the high-pass cutoff frequency on IED. The cutoff frequency, defined for $|H(z)| = -3$dB (in linear scale: $|H(z)| = \frac{\sqrt{2}}{2}$), is:

$$f_c = \frac{1}{\pi\ \mathrm{IED}} \sin^{-1}\left(\frac{\sqrt{2}}{4}\right) \approx \frac{0.115}{\mathrm{IED}} \tag{13}$$

Equation (13) indicates that surface potentials with spatial periods wider than $\sim$ 8.7 IED would be filtered ($|H(z)|_{f>8.7\mathrm{IED}} < -3$dB) in EMGs detected by two electrodes in bipolar montages. That is, IED determines the sensitivity of bipolar recordings. For the sake of clarity, electric sources generating a surface potential with spatial periods wider than 87 mm, 174 mm and 261 mm in the center-to-center direction of bipolar electrodes would be attenuated by 3 dB in recordings respectively taken with 10 mm, 20 mm and 30 mm IEDs. Biophysically, signal properties are inherently linked with the IED used.

To overcome these biophysical constraints, we must address the second technical issue of concern for the definition of the minimal IED maximizing EMG sensitivity: What is the lowest spatial frequency defining the bandwidth of the surface representation of electric sources within the target muscle? We believe a generally valid answer to this question is elusive owing to the multitude of factors affecting the surface representation of action potentials [9], [11]. Some of these factors—e.g., the position of the fibers that are excited within a single motor unit-cannot be controlled for experimentally, which likely justifies the absence of *in vivo* studies attempting to characterize the spatial bandwidth of electric sources in a single, target muscle. Therefore, here we i) assess the relevance of the problem, showing how IED may critically affect the sensitivity of bipolar recordings and ii) propose a possibly generalizable, valid solution to the problem: allowing the IED to vary between muscles and subjects to facilitate the representativeness of bipolar, surface EMGs. Our reasoning is grounded on the biophysical association between IED and the high-pass filtering of surface potentials (13) and on previous evidence establishing a dependence of the optimal IED on muscle size [14], [15].

[2]As defined here, $H(z)$ implies the position of one of the electrodes, $V_{ei+1+d[k]}$ in (1), projects onto the origin of the axis defining the center-to-center direction of the two electrodes. The reference system used to represent $H(z)$ does not affect the validity of (12) and (13).

B. The Dependence of Bipolar EMG Descriptors on the IED

The results reported here support our hypothesis: bipolar EMGs detected with greater IEDs are more sensitive to muscle excitation. Evidence favoring the increased sensitivity of EMGs detected with greater IEDs, as recently documented in other studies [3], [14], can be summarized by the changes we observed in the EMG surrogates for the timing and degree of excitation of the two muscles assessed. As IED increased, from 5 mm to 50 mm: (i) the variability of EMG-Torque onset decreased, with IEDs greater than 30 mm providing onset values converging to tenths of a ms earlier in relation to the average torque onset (Fig. 4); (ii) SNR values increased, saturating at 30 mm and 90 mm IEDs (IED $= 3\tau^{IED}$, evaluated at 10% MVC) in the biceps brachii and gastrocnemius, respectively (Fig. 5D). We contend that these findings follow from theory.

In our view, the mechanistic basis of the IED-sensitivity relationship is presumably determined by the high-pass cutoff frequency of the bipolar filter (12) and the filtering effect of the conduction volume [15], [35]-that is, the tissue between the skin and the electric sources. By increasing the IED, the high-pass cutoff shifts to lower frequencies, increasing the contribution of the energy of sources with wider spatial periods to the surface EMG. The increased sensitivity provided by greater IEDs, as reported elsewhere for EMG amplitude values [4], [16], [36] and as reported here for the increased SNR values (Fig. 5), corroborates this view. Although the monotonic increase in SNR with contraction level is not surprising (Fig. 5C), owing to the putative EMG-force relationship [36], the saturation of SNR values with IED may be (Fig. 5A-B). This apparent contrast may arise if one incorrectly expects the amplitude of bipolar EMGs to depend only on the IED. In addition to the IED effect, the amplitude of the detected signal is limited by the low-pass filtering effect of the conduction volume. Modelling the filtering effect of the conduction volume is not as straightforward as that of IED (12). Nevertheless, it is well established that greater distances between the source and the skin yield lower and wider electrical potential surface distributions [18], [35]. Therefore, increasing IED improves the surface representation of distant sources, provided that they are sufficiently close to the pair of electrodes for their amplitude to exceed the noise level inherent in any acquisition system. The surface representation of action potentials is ultimately limited by both IED and the distance of the excited fibers to the skin (Fig. 3).

Although sensitivity improves with greater IEDs, the question remains whether the sources driving this sensitivity are located within the target muscle or the surrounding muscles. In other words, is this sensitivity sacrificing specificity? At least for IEDs shorter than 30 mm, we believe there is limited specificity trade-off. More specifically, the unreasonably high onset values with IEDs$<$30 mm suggest poor sensitivity in these conditions-we are not capturing sources from the target muscle. On occasion-and as reported in [27] for 10mm IED-EMG onset was delayed with respect to torque onset (i.e., negative electromechanical delay; Fig. 1 and 3). Only for IEDs$\geq$30 mm did EMG onsets reliably precede the average torque onset. These biophysically feasible outcomes were obtained for both muscles in 18 out of the 19 subjects, and their estimates approach the expected range of electromechanical delays ($<$70ms; [37], [38]). Therefore, it seems likely that sources within the target muscle have been filtered out (13) by the differential, bipolar electrodes with IEDs shorter than 30 mm. These findings substantiate the applied relevance of Type II errors in surface EMG research [9]; that is, the lack of EMG's sensitivity to excitation within the target muscle. Implicit in our reasoning is the well-grounded notion that both BB and GM, being at their rest length, were indeed excited and contributed to the rise in joint torque during the isometric contractions. An alternative explanation for the spuriously delayed onset values would be that, for all subjects tested, joint torque was entirely accounted for by distant muscles, which representation in surfaced EMG emerged only for IEDs$\geq$30 mm. In our view, and based on highly selective, intramuscular EMGs [39], this event is unlikely and a reasonable explanation would be needed to justify the lack of BB and GM contribution to the rise in joint torque in the subjects tested.

As anticipated in the Introduction, we do not intend to misconstrue the importance of Type I error (crosstalk). Rather, our goal is for readers to ponder and weigh the relevance of both Type I and Type II errors when devising experiments and interpreting data, bearing in mind that surface EMGs detected with IEDs shorter than 30mm from the biceps brachii or the gastrocnemius muscle are more prone to Type II error. We are aware that eliminating both errors is impractical; indeed, there is necessarily a mathematical tradeoff between false positives and false negatives. This, in turn, forces EMG users to minimize and balance their errors according to the circumstances. For example, short IEDs would yield poor sensitivity in subjects with remarkably thick subcutaneous tissues [40] or in conditions imposing relative changes between the bipolar electrodes and the target, muscle volume [16]. Similarly, when the excitation of surrounding muscles is limited to 30% of their maximum and that of the target muscle is lower than 50% of its maximal, the degree of crosstalk (Type I) has been shown to increase by up to roughly 5% when IED increased from 10 mm to 40 mm [4]. In such an instance, one may suggest the 5% increased chance of crosstalk justifies the collection of more sensitive EMGs, particularly for studies assessing the timing of muscle excitation or the EMG-force relationship. Conversely, when assessing muscle cocontraction in patients with pathology who have difficulty in selectively recruiting the agonist muscle [41], one may wish to strive for a lowest possible Type I error. In general, when the ratio of excitation between surrounding and target muscles is expectedly small, we feel confident recommending users to sample bipolar EMGs with 30 mm IED for the BB and GM. When using smaller IEDs, researchers should acknowledge the possibility of Type II errors in their results.

C. Proposing a Generalizable Valid Optimal IED for Single Bipolar, Surface EMG Recordings

Two issues are of concern regarding the generalization of our results. The first is to different types of electrodes and the second is to different muscles. The high-pass filtering effect

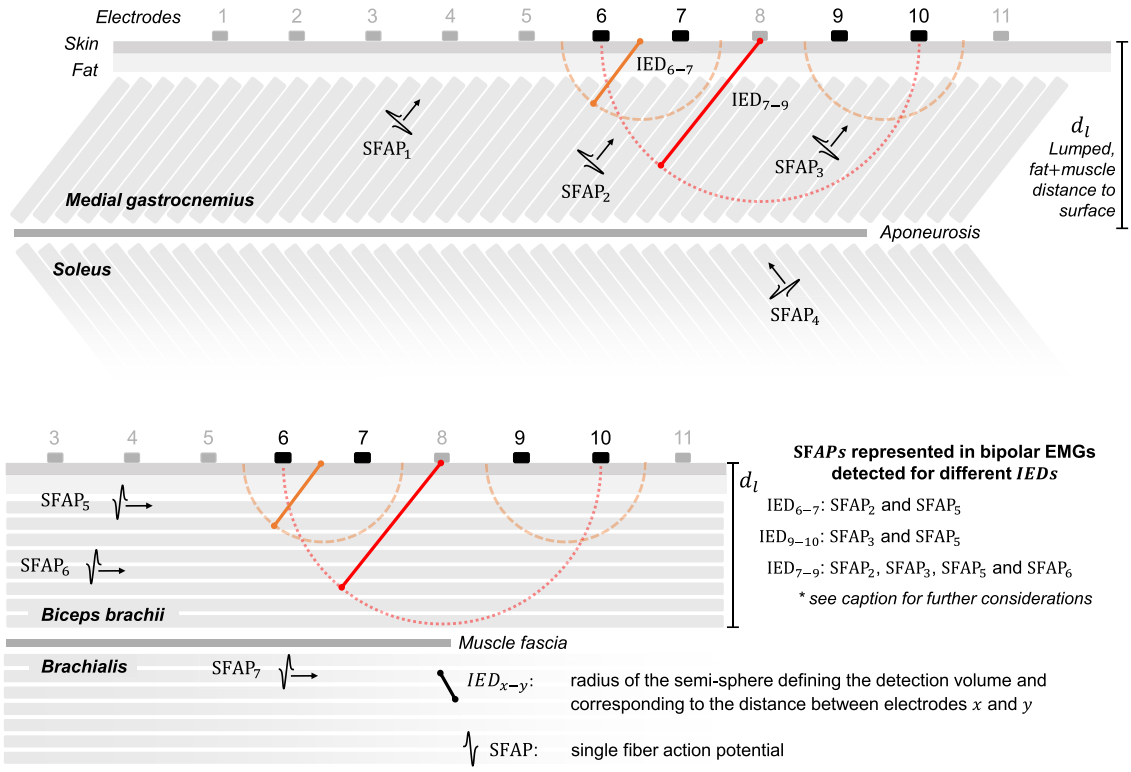

Fig. 6. The trade-off between sensitivity-specificity is affected by IED. Semi-circles were drawn according to the one-to-one relationship between the amplitude of surface potential and the IED radial distance to the center of the pair of electrodes, advanced by [15] and conservatively corroborated by our findings with intramuscular and high-density surface EMGs [14]. For the example shown, the maximal possible IED that is capable of avoiding Type I errors (crosstalk) and minimizing Type II errors in both muscles is provided by any pair of consecutive, even or odd electrodes (e.g. $IED_{\mathbf{7-9}}$). However, action potentials in GM distal fibers would still be missed (e.g. $SFAP_{\mathbf{1}}$). Further increasing IED would reduce Type II errors in GM, but at the cost of increasing Type I errors (representation of $SFAP_{\mathbf{4}}$, and $SFAP_{\mathbf{7}}$ in BB). Multiple bipolar electrodes are thus necessary if Type II error is to be avoided for GM. We proposed IED be defined according to individuals and muscles, being scaled to the lumped, fat+muscle distance to the surface: IED $< d_l$. The possible consequences of using shorter distances are presented in Fig. 4-5.

associated with the spatial, differential operation of bipolar recording (12) stands regardless of the electrode dimension and shape. Indeed, as experimentally tested by [4], bipolar EMGs detected with both rectangular (1 × 10 mm) and disc (10 mm diameter) electrodes depend equally on IED. Regarding electrode dimension, the only systematic account we identified was a thorough, in silico study [21]. Their results confirmed the increased energy of lower spatial frequencies as electrode size increases, regardless of their shape: the cutoff frequency of the low pass filtering associated with electrodes of finite size decreases as the electrode size increases (cf. Fig. 2 in [21]). Our simulation results (Fig. 3) corroborate this view. In brief, from theory [21], the Fourier transform of the detected potential results from the product between the:

1) Fourier transform of the potential over the skin;
2) Electrode transfer function, which depends on the electrode size and shape [19], [21]
3) Spatial filter transfer function, which depends on the IED (our equations; [31])

Point (2) posits a low-pass filter, with lower cutoff frequencies for larger electrodes (cf. Fig. 2 in [21]). Point (3) posits a high-pass filter, with higher cutoff frequencies for shorter IEDs (cf Fig. 4b in [31] and (13)). Considering the product of both, it appears evident that the effect of IED reported is invariant to electrode size. Indeed, Fig. 3 shows that, apart from an offset, the relationship between EMG amplitude and IED does not depend on electrode size, regardless of the muscle architecture and location of fibers simulated: larger electrodes appear detrimental, providing lower EMGs, when the excited fibers are closer to the center of the pair of electrodes (Fig. 3). Overall, therefore, current evidence suggests the considerations made here apply to different types of electrodes.

Generalizing our findings to different muscles is more challenging. At a first glance, we would feel tempted to advocate the sampling of EMG with multiple electrodes from a single muscle as the most appropriate procedure for contending with Type II error. However, studies relying on single bipolar recordings for each muscle would not benefit from this recommendation. Most importantly, even though high-density technology provides a more accurate representation of muscle excitation in surface EMGs [9], our current results contest this possibility as a generally valid solution. Our reasoning is schematically illustrated in Fig. 6. In a parasagittal section, for in-depth pinnate muscles, the sensitivity of bipolar surface EMGs is a function of the number of pairs of electrodes and of their IED. As shown in Fig. 6A, optimizing both sensitivity and specificity for in-depth pinnate muscles would either require many pairs of closely spaced electrodes or a few pairs of widely spaced electrodes. It is therefore not surprising that SNR values for gastrocnemius saturate at a much

greater IED (90 mm; $3\tau^{IED}$) when compared to the biceps brachii (Fig. 5). For skin-parallel fibered muscles, sensitivity of bipolar EMGs is mainly a function of IED. Multiple pairs of closely spaced electrodes placed parallel to the muscle fibers would only provide a delayed representation of the same sources, propagating along the fibers. The detection of deeper sources would demand increasing IED (Fig. 6B). Based on Fig. 3-5, using grids of electrodes with IEDs greater than those often considered (IED $\leq$ 10mm) would be warranted if physiologically valid inferences were to be drawn from bipolar EMGs collected from skin-parallel fibered muscles.

While manufacturers offer the possibility of sampling bipolar EMGs with different montages, from fixed to variable IEDs, from small (less than 0.5 cm2 contact area) to large, disk electrodes (up to roughly 5 cm^2 contact area), existing evidence guiding users on which IED to use when and why is elusive. The reference most often relied upon in this regard likely stems from the European Concerted action SENIAM, born when specificity was the only, apparent concern in surface EMG-for the two muscles considered in this study, a fixed 20 mm IED is recommended [42]. From our current results, it seems plausible to propose the definition of IED on an individual and muscle basis.

The key question here is: How can we define an optimal, generally valid IED for a single, bipolar EMG recording? Fig. 6 shows that indeliberately increasing IED is not a candidate answer. If IED increases too much, specificity is sacrificed. The difficulty here is understanding when IED is too large. So far, to our knowledge, the most illuminating evidence relating IED to the sampling of action potentials from fibers located at different distances from the electrodes has been provided by [15]. They rigorously demonstrated the surface amplitude of a single action potential depends on the radial distance between its source, the excited fiber, and the center of the pair of electrodes. When the fiber is located one IED unit away from the electrodes, the amplitude of its action potential is roughly 10% of the amplitude of an action potential generated in fibers located 0.2 IED units from the electrodes. This attenuating effect of distance is accentuated when using the root mean square value to assess the signal amplitude (cf. their Fig. 5). With intramuscular electrodes and an array of surface electrodes, we experimentally substantiated the simulation results reported by [15]. We specifically observed that gastrocnemius specificity was sacrificed for IEDs greater than roughly 1.5-times the lumped (fat + gastrocnemius) thickness (cf. Fig. 7A in [14]). Collectively, this evidence seems to undermine any attempt in recommending a generally valid, optimal IED. The variance between individuals and between muscles must be considered when optimizing the IED in single, bipolar studies. When methodologically possible, we recommend scaling IED according to the lumped, fat-muscle distance between the center of the pair of electrodes and the closest muscle border, using a one-to-one scale. Following the discussion in this section, our recommendation is expected to hold for both large and small muscles, providing the electrode size allows for sampling from the target muscle with a one-to-one scaling. Examples are provided in Fig. 6 for the sake of clarity. When figures for the lumped distance are not available, results should be discussed in the light of Type I or Type II error, in particular when IED is respectively larger or smaller than the expected fat-muscle distance to electrodes.

V. Conclusion

In this study, we showed that IED imposes a biophysical constraint on the sensitivity of bipolar surface EMGs. We further documented how IED critically affects the physiological content of the surface EMG, with short IEDs inflating Type II errors. Low quality surface EMGs provide physiologically unreasonable estimates of excitation onset and are less sensitive to changes in muscle excitation, which are the consequences of an inflated Type II error for both gastrocnemius and biceps brachii muscles. Decreasing Type II error in surface EMG necessitates appropriate electrode spacing, based on a muscle-specific approach rather than on the use of a single, generally valid IED. Specifically, we propose that IEDs be scaled according to the distance between the muscle border and the center of the pair of electrodes (Fig. 6). We expect these results to increase awareness of the physiological importance of Type II errors in single bipolar studies: i) sampling surface EMGs from large pinnate muscles, as mitigation of Type II errors in pinnate muscles may demand multiple recording points (Fig. 6) and; ii) using small IEDs for gastrocnemius, biceps brachii and, presumably, other similarly large muscles, which may yield inferences that are constrained to a small, unrepresentative volume.

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