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Surface EMG: The issue of electrode location

L. Mesin [a], R. Merletti [a,*], A. Rainoldi [b]

[a] *Laboratory for Engineering of the Neuromuscular System (LISiN), Department of Electronics, Politecnico di Torino, Italy*
[b] *Motor Science Research Center, School of Motor Sciences (SUISM), Università di Torino, Torino, Italy*

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Abstract

This paper contributes to clarifying the conditions under which electrode position for surface EMG detection is critical and leads to estimates of EMG variables that are different from those obtained in other nearby locations. Whereas a number of previous works outline the need to avoid the innervation zone (or the muscle belly), many authors place electrodes in the central part or bulge of the muscle of interest where the innervation zone is likely to be. Computer simulations are presented to explain the effect of the innervation zone on amplitude, frequency and conduction velocity estimates from the signal and the need to avoid placing electrodes near it. Experimental signals recorded from some superficial muscles of the limbs and trunk (abductor pollicis brevis, flexor pollicis brevis, biceps, upper trapezius, vastus medialis, vastus lateralis) were processed providing support for the findings obtained from simulations. The use of multichannel techniques is recommended to estimate the location of the innervation zone and to properly choose the optimal position of the detection point(s) allowing meaningful estimates of EMG variables during movement analysis.

Keywords: Surface electromyography; EMG; Electrode location; Electrode position; Electrode arrays

1. Introduction

During the last 30 years the effect of electrode location on estimates of conduction velocity (CV), amplitude and spectral variables of the surface EMG has been addressed in a number of methodological and clinical publications (Castroflorio et al., 2005; Cote and Mathieu, 2000; Falla et al., 2002; Fuglevand et al., 1992; Hogrel et al., 1998; Kiryu et al., 1996; Li and Sakamoto, 1996; Lynn et al., 1978; Mercer et al., 2006; Roy et al., 1986; Zipp, 1978; among many others) considering muscles (or groups of muscles) ranging from the masticatory muscles to the muscles of the shoulder, of the arm and leg. The most significant standardization effort took place in 1997–1999 within the European Project on "Surface EMG for Non Invasive Assessment of Muscles" (SENIAM) [www.seniam.org] where a detailed analysis of literature was presented for a number of muscles. Three strategies for placement of an electrode pair resulted from this analysis as the most used and are reported below (only data concerning the biceps brachii muscle are reported as an example):

– on the center or on the most prominent bulge of the muscle belly (in 10 out of 21 publications);
– somewhere between the innervation zone (IZ) and the distal tendon (in 6 out of 21 publications);
– on the motor point (in 1 out of 21 publications).

In 4 out of the 21 reviewed publications the electrode location was either not mentioned or was unclear, indicating the little attention paid by authors and reviewers to this issue.

The orientation of the detection system (usually constituted by an electrode pair) with respect to the direction

Abbreviations: CV, muscle fiber conduction velocity; DD, double differential; IED, interelectrode distance; IZ, innervation zone; MU, motor unit; MUAP, motor unit action potential; NMJ, neuromuscular junction; SD, single differential.

* Corresponding author. Tel.: +39 011 4330476; fax: +39 011 4330404.
E-mail address: roberto.merletti@polito.it (R. Merletti).

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of muscle fibers was rarely mentioned. The SENIAM recommendation of avoiding placing electrodes over the IZ was based on criteria that were only partially documented within the project itself.

The objective of this work is the discussion and updating of considerations and criteria for surface EMG electrode location on the basis of both computer simulations and experimental data.

2. Methods

2.1. Simulations

Computer simulation of the sources and of the volume conductor produces the potential distribution on the surface of the skin. The model described in Farina and Merletti (2001) has been used to compute the single differential (SD) potential present under each electrode pair on the skin (skin thickness 1 mm, conductivity 0.022 S/m; fat thickness 3 mm, conductivity 0.04 S/m; semi-infinite muscle layer with longitudinal conductivity 0.4 S/m and transversal conductivity 0.09 S/m). Single fibers of semi-length equal to 60 mm have been simulated and their contributions have been added to generate the motor unit action potential (MUAP). The contributions of three motor units (MU) have been added to simulate a simple interference EMG signal. For all muscle fibers conduction velocity (CV) is 4 m/s.

2.2. Experimental data

Surface EMG signals from the muscles listed in Table 1 were considered. Some of the experimental data have been already discussed in previous works of our group (see references in Table 1), which the reader can refer to for a detailed description of the experimental set-up. Surface EMG signals were recorded in SD configuration during isometric contractions. Two contraction levels expressed in terms of percentage of maximal voluntary contraction (MVC) were considered (except for vastus lateralis and medialis), one level was low (lower than or equal to 30% MVC), the other high (equal to or higher than 50% MVC). In the case of the biceps brachii, the dominant arm was fixed in a brace with angle of the elbow 105° (full elbow extension = 180°). Linear arrays aligned with muscle fibers were used to detect surface EMG except for upper trapezius and biceps brachii, in which case two-dimensional arrays were used in the acquisition of data. For biceps brachii the same detection system described in Troiano et al. (2008) was used (13 rows by 5 column aligned with the muscle fibers), but only the central column was considered.

2.3. Signal processing

The position of the main IZ was identified by visual analysis of the EMG signals as the channel under which the phase inversion of the MUAPs detected in SD mode occurred, and where the MUAPs began to propagate in two opposite directions. Amplitude and spectral variables were estimated from SD channels, CV was estimated from pairs of double differential (DD) channels with a maximum likelihood algorithm. The EMG variables averaged rectified value (ARV), mean spectral frequency (MNF) and CV were computed for the channel corresponding to IZ, the three distal channels and the three proximal channels (two distal and two proximal channels in the case of vastus medialis and lateralis), on eight 0.5 s long adjacent epochs (at the beginning of the contraction, to reduce the effect of fatigue), for each signal. The values of the variables obtained in eight different epochs were then averaged (considering only CV values between 1 and 10 m/s and discarding up to four outliers, i.e. values distant from the mean more than twice the standard deviation). Thus, for each signal array (i.e., fixing the muscle, the subject, and the force level), seven average values (or five, in the case of vastus medialis and lateralis) were obtained for each of the variables under consideration (ARV, MNF and CV) corresponding to the seven (or five) geometrical locations. The signals were processed only if three (or two) channels distal and proximal with respect to the IZ were available, so that seven (or five) locations could be considered (only some of the subjects under study in the quoted papers were thus considered herein). Table 1 provides the number of the subjects considered for each muscle.

In order to compare signals from different subjects, the time averaged variables corresponding to each signal were linearly scaled so that the maximum value (across different channels for each subject) was assigned the value 1 and the minimum value was assigned to 0. The adopted scaling equation was

$$x_{\mathrm{norm}} = \frac{x - x_{\mathrm{m}}}{x_{\mathrm{M}} - x_{\mathrm{m}}}$$

Table 1
List of muscle from which experimental data have been collected

Muscle	IED (mm)	Force levels (% MVC)	Number of subjects	Reference
Abductor pollicis brevis	2.5	10 and 50	5	Rainoldi et al. (2008a)
Flexor pollicis brevis	2.5	10 and 50	5	Rainoldi et al. (2008a)
Biceps brachii	8	30 and 70	8	This work
Upper trapezius	8	20 and 80	6	Troiano et al. (2008)
Vastus lateralis	5	90	10	Rainoldi et al. (2008b)
Vastus medialis	5	90	10	Rainoldi et al. (2008b)

where x is the current value of the considered variable, x_{norm} is its normalized value, x_m and x_M are the minimum and maximum value observed along the array (±3 inter-electrode distance (IED) or ±2 IED for vastus medialis and lateralis).

One way non-parametric Friedman ANOVA for repeated measures was performed to assess if electrodes/channels (that is the positions over the muscles) play a role as factor in the distribution of variance. When the test indicated significant variations (significance level always set to $p < 0.05$), pair-wise comparisons were performed with the Dunns post-hoc test looking for statistically significant differences among pairs of averaged variables from individual channels.

3. Results

3.1. Simulations

Let us first consider a single muscle fiber parallel to the skin. Two point electrodes placed on the skin symmetrically with respect to the neuromuscular junction (NMJ) and along the fiber direction detect a differential signal that is obviously zero, as shown in Fig. 1, due to specularity of the propagating signals.

Similar results are obtained in the case of a single MU with fibers parallel to the skin and NMJs uniformly distributed within the IZ. A sharp minimum of EMG amplitude is obtained when the electrodes are over or near the IZ. The signal amplitude, but not its global pattern, is affected by the averaging effect introduced by the scatter of the NMJs within the IZ and by the physical size of the electrodes. Fig. 2 shows the results of computer simulations (Farina and Merletti, 2001) of a single MU (IZ 10 mm wide) whose MUAP is detected by a differential electrode pair with IED of 5, 10 or 20 mm in different sets of simulations. MNF and CV (estimated from pairs of DD signals) are also shown. The detection point (center point between the two electrodes in the case of ARV and MNF; center point between the two DD detection systems used in the case of CV) is moved from the center of the IZ to the tendon junctions 60 mm away, in steps of 1 mm. Both point electrodes and rectangular electrodes are simulated.

The zone of stable EMG amplitude is relatively narrow, especially in the case of IED comparable with the fiber semi-length, and 25–40 mm away from the IZ. Only in such a narrow region, estimation of ARV, MNF and CV is reliable.

In the case of interference signal generated by many MUs innervated in approximately the same location, the ARV minimum is less sharp depending on the scatter of the IZs. Displacements of the electrodes of the order of 10 mm result in changes of amplitudes that depend on the width and spread of the IZs of the active MUs that are within the detection volume of the electrode system. Fig. 3 shows this concept for a pair of rectangular electrodes aligned with the fiber direction and depicts a more complex and realistic situation than that shown in Fig. 2,

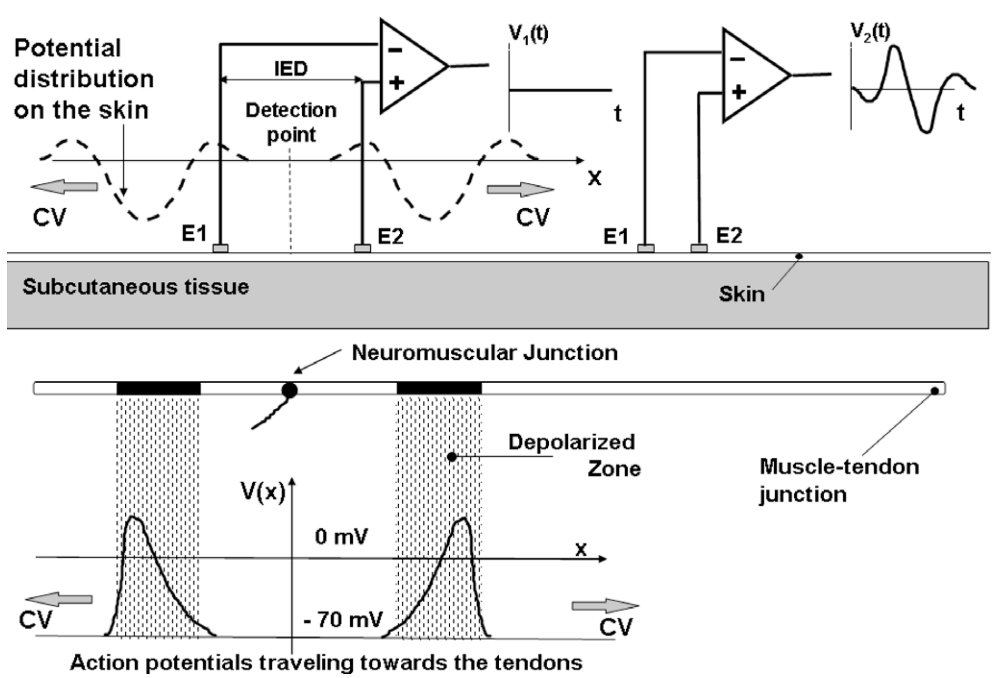

Fig. 1. Surface EMG signal generated by a single muscle fiber. Two spatial distributions of potentials, each associated to a depolarized zone, propagate from the neuromuscular junction to the tendon endings. Two electrodes placed symmetrically over the neuromuscular junction detect identical signals whose difference is zero, while two electrodes placed on one side of the junction detect a non zero differential signal. $V(x)$ is the transmembrane potential. IED is the interelectrode distance, the detection point is the center point of the electrode pair. $V_1(t)$ and $V_2(t)$ are the outputs of the differential amplifiers.

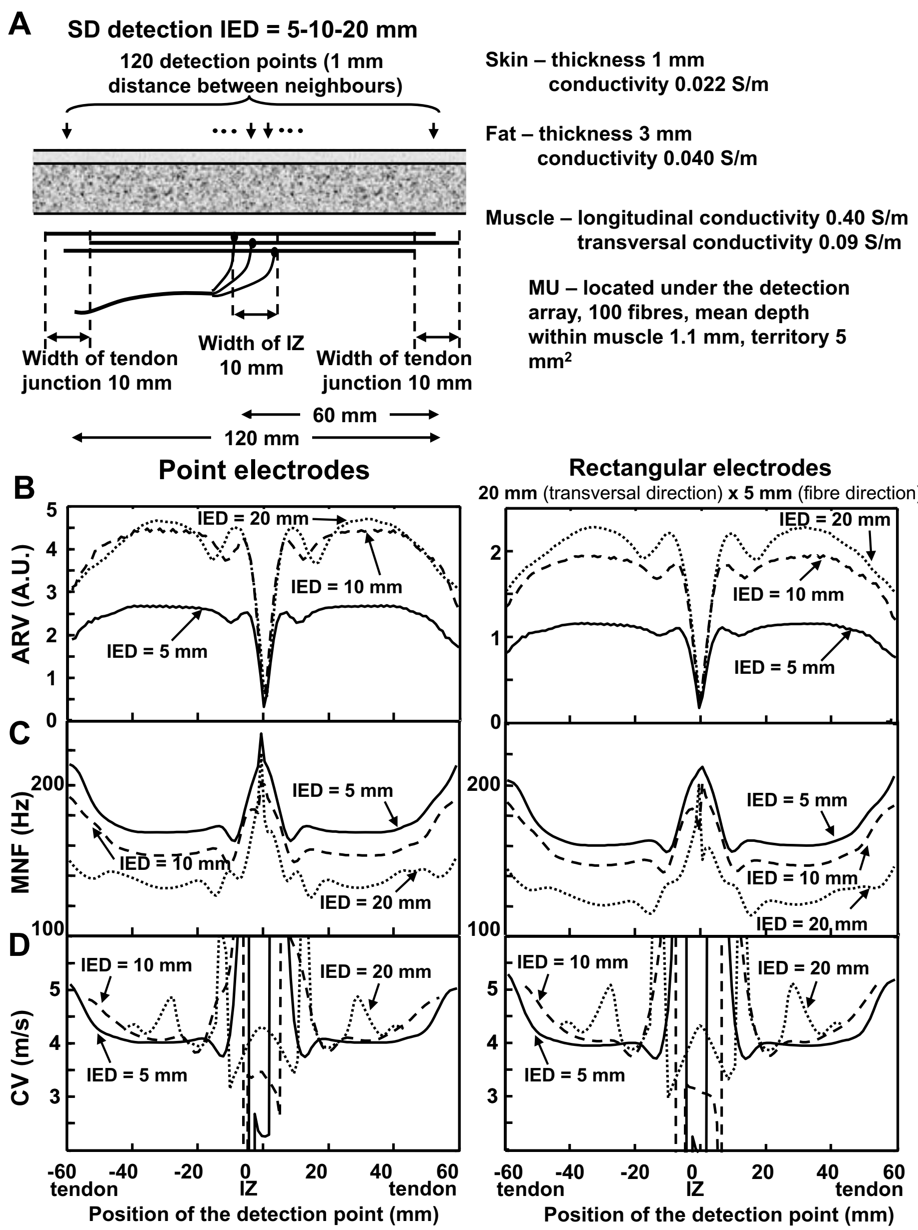

Fig. 2. (A) Simulation of action potentials generated by a motor unit and detected by point or rectangular (20 × 5 mm) electrodes for different interelectrode distances (IED) and detection point locations. (B) Average rectified value (ARV), and (C) mean frequency (MNF) of the single differential (SD) EMG signal detected between two electrodes (5, 10 or 20 mm apart) moved over the surface of the skin above a single 120 mm long motor unit in steps of 1 mm. (D) Muscle fiber conduction velocity (CV) estimated from a pair of double differential (DD) signals (with IED 5, 10 or 20 mm) moved over the surface of the skin above the same motor unit in steps of 1 mm. The neuromuscular junctions of the individual fibers are uniformly distributed over the innervation zone (IZ) which is 10 mm wide. For IED = 20 mm, the length of the four electrode array required to generate the two DD signals for CV estimation equals the distance between the IZ and the tendon.

where three identical MUs, placed at the same depth next to each other, have the respective IZs shifted in space by 5, 10 or 20 mm.

3.2. Experimental data

Fig. 4 shows an example of processing procedure applied to experimental signals from upper trapezius muscle (contraction level 80% MVC, IED = 8 mm, six subjects considered). IZ is determined by visual analysis and the channels corresponding to IZ, the three distal channels and the three proximal channels are selected (Fig. 4A). ARV, MNF and CV are estimated for the seven channels considered, for each of the six subjects (Fig. 4B on the left). These variables are normalized (Fig. 4B on the right). Thus, at this point, all data are aligned and are scaled in amplitude, so that data from different subjects can be compared. Statistical analysis is then performed on normalized and aligned data (Fig. 4C).

Non parametric ANOVA indicated statistically significant dependence of ARV and MNF from the detection point for all considered muscles (except for abductor polli-

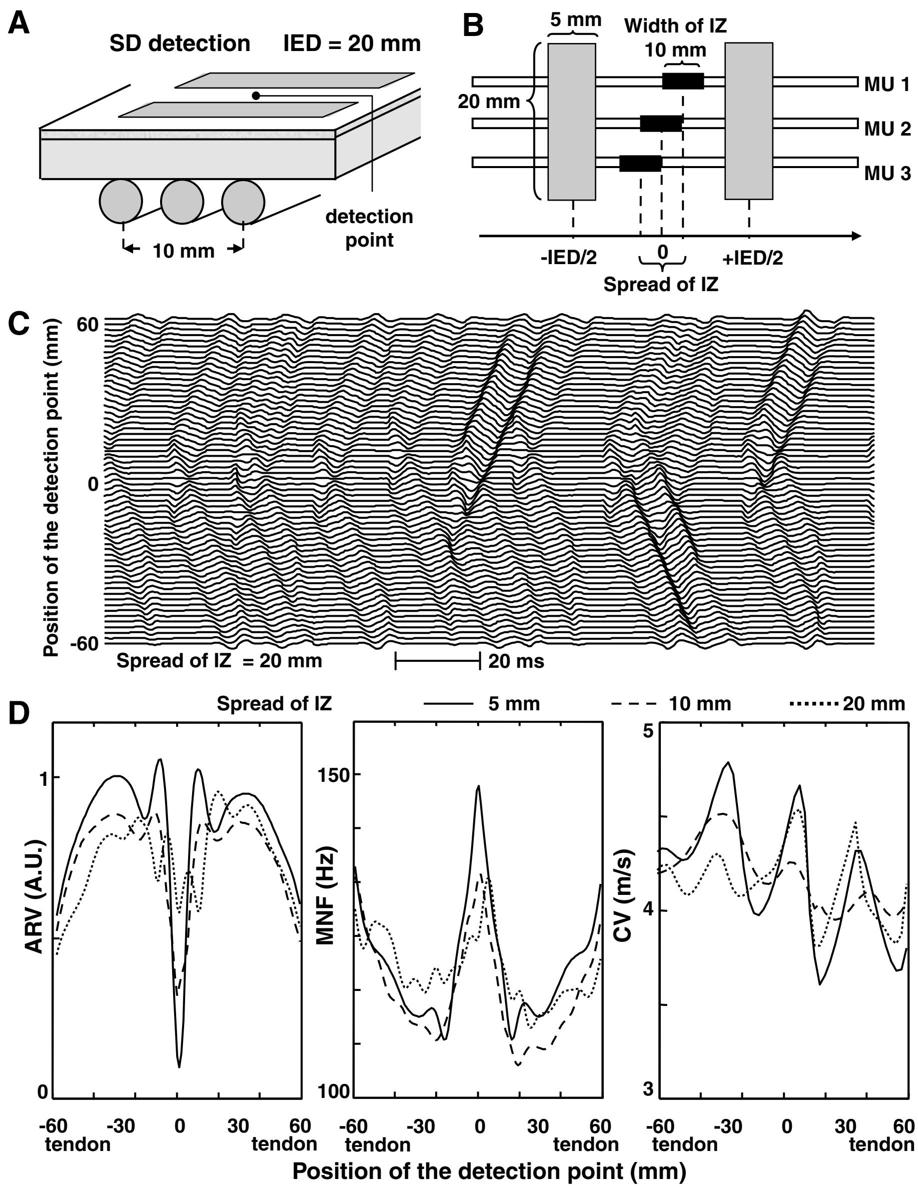

Fig. 3. (A) Geometry of three identical simulated motor units with different IZs and the same CV = 4 m/s. (B) Geometry of the electrode pair (rectangular electrodes, 20 × 5 mm, IED = 20 mm) and of the IZ. (C) Examples of simulated raw signals for different locations of the detection point. (D) Plots of ARV, MNF and CV of the EMG for different spread of the three IZ (each is 10 mm wide) and different locations of the detection point from one tendon to the other.

cis brevis (MNF at 10 and 50% MVC and ARV at 50% MVC). CV was statistically dependent on the detection point only for signals detected over the upper trapezius muscle. Dependence of EMG variables on channel position was statistically significant with $p < 0.01$ in the case of all muscles except for abductor pollicis brevis and flexor pollicis brevis (for the latter significance was at $p < 0.05$). Table 2 shows the results of the post-hoc test. The values of the variables ARV, MNF and CV estimated from signals detected in different channels were compared to the values estimated from the signals detected over the IZ. The minimum distance (in terms of number of channels) from the IZ for which the post-hoc test disclosed statistically significant difference is provided, for both proximal and distal directions.

4. Discussion and conclusions

A differential montage with both electrodes placed on one side of the IZ detects mostly unidirectionally propagating MUAPs. In this case, estimates of EMG amplitude, spectral variables and CV are less affected by minor electrode displacement as well as by the potentials propagating in the opposite direction. Multichannel montage (with $N > 2$) must be used in this region if muscle fiber CV is to be estimated and if limited sensitivities of amplitude and spectral estimates with respect to electrode displacement are desired (Farina et al., 2002a,b; Hogrel et al., 1998; Roy et al., 1986).

Simulations of simple situations are provided in Figs. 2 and 3. Experimental data were investigated during isomet-

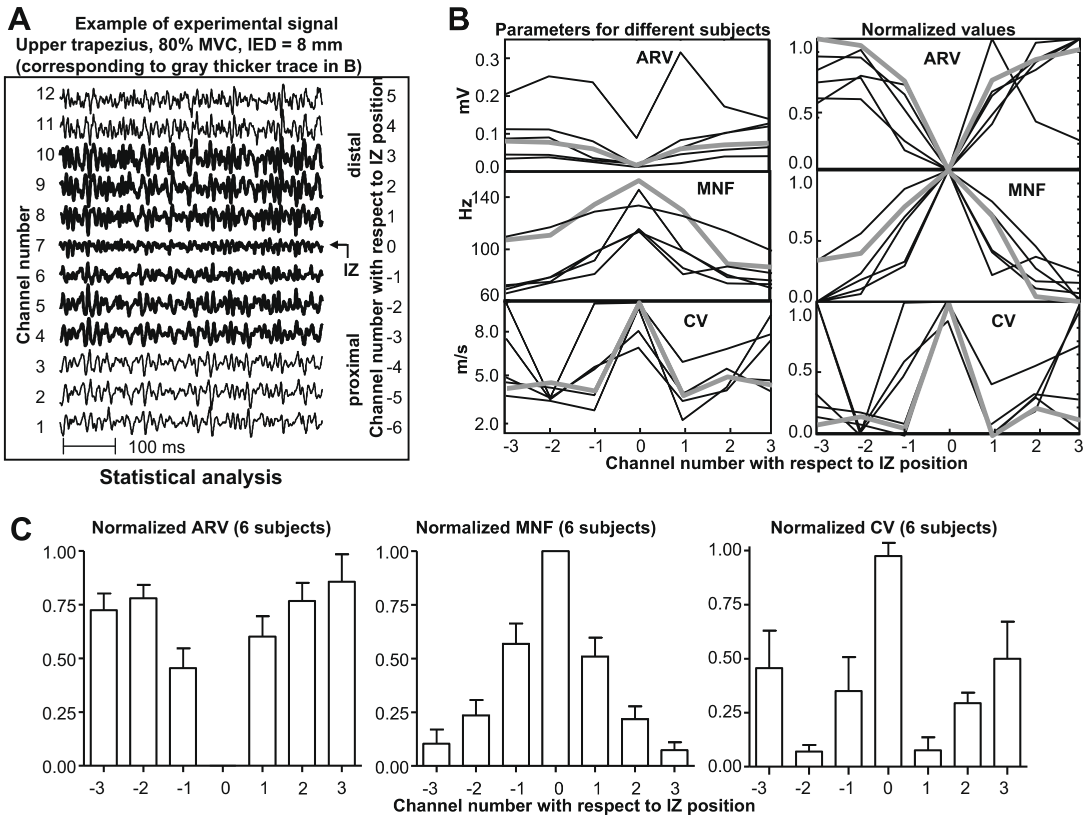

Fig. 4. (A) Example of experimental signals recorded during the contraction at 80% MVC of the upper trapezius muscle. IZ determined by visual analysis and the channels selected for subsequent processing are indicated. (B) ARV, MNF and CV (dimensional and normalized values) estimated for the seven channels considered, for each of the six subjects. (C) Statistical analysis performed on the data normalized with respect to the minimal and maximal values and scaled in the range 0–1. Similar results were obtained from other muscles (abductor pollicis brevis, flexor pollicis brevis, biceps brachii, vastus medialis, vastus lateralis).

Table 2
Electrode locations for which the differences, with respect to the values over the IZ, were found to be statistically significant after the post-hoc test

Muscle	Force levels (% MVC)	Minimum distance for statistical difference ($p < 0.05$)		
		ARV	MNF	CV
Abductor pollicis brevis	10	3 Channels (proximal)	NS	NS
($N = 5$)	50	NS	NS	NS
Flexor pollicis brevis ($N = 5$)	10	2 Channels (proximal)	3 Channels (distal)	NS
	50	2 Channels (proximal)	3 Channels (distal and proximal)	NS
Biceps brachii ($N = 8$)	30	2 Channels (distal)	3 Channels (distal and proximal)	NS
	70	2 Channels (distal)	3 Channels (distal)	NS
Upper trapezius ($N = 6$)	20	3 Channels (distal or proximal)	2 Channels (distal) 3 Channels (proximal)	1 Channel (distal) 2 Channels (proximal)
	80	2 Channels (distal or proximal)	3 Channels (distal or proximal)	1 Channel (distal) 2 Channels (proximal)
Vastus lateralis ($N = 10$)	90	2 Channels (distal)	2 Channels (distal)	NS
Vastus medialis ($N = 10$)	90	2 Channels (distal)	2 Channels (distal)	NS

NS: not significant difference, N: number of subjects.

ric non fatiguing contractions, confirming the results obtained from the simple simulations considered. The superficial muscles of the hand, the limbs and the trunk listed in Tables 1 and 2 were studied for which fibers can be considered rectilinear, parallel to each other and to the skin surface within a first approximation. Potentials propagating from IZ to tendons (with small distortion in shape) could be detected.

Fiber inclination with respect to the skin surface and other geometrical factors which vary from muscle to muscle, person to person (Farina et al., 2002b; Rainoldi et al., 2000, 2004), and for the same muscle during shortening

(Schulte et al., 2004) or fatigue conditions can strongly influence the detected signal. Furthermore, there are (relatively infrequent) cases of muscles with multiple IZs lacking an area of unidirectional propagation. In these cases (not shown) estimation of global CV and EMG amplitude may indeed be affected by large fluctuations in space and it may be necessary to resort to electrode arrays from which the individual MUAPs may be extracted, classified and analyzed (Gazzoni et al., 2004; Holobar and Zazula, 2004) (some fluctuation in CV estimation is also shown for the simple simulations shown in Fig. 3, where local maxima of CV estimation are found over each of the three simulated IZs). The increasing use of mathematical models (Farina and Merletti, 2001; Hogrel et al., 1998) and of electrode arrays and multichannel amplifiers (Merletti et al., 2001, 2003; Ostlund et al., 2007; Saitou et al., 2000) at the clinical research level (Beck et al., 2008; Campanini et al., 2007; Cescon et al., 2008) is providing insight in these situations.

The basic principle depicted in Figs. 1 and 2 explains and confirms the clinical observation that the amplitude of the EMG signal detected with a pair of electrodes has a minimum over the IZ (Beck et al., 2008; Rainoldi et al., 2000). Furthermore, Fig. 2 shows that reliable estimates of MNF and CV can be obtained only in regions where EMG amplitude is stable to small displacements along fiber direction. This result is also supported by the experimental signals considered in this study. To compare data from different subjects, a normalization procedure was used. The considered normalization forces the data of each subject to cover the full range 0–1, loosing information about relative changes of the variables across channels, but enhances common patterns. Indeed, statistically significant dependence of EMG variables from the detection point was found for all the muscles under study.

When the MUs active in the detection volume have IZs scattered in space or when bipolar detection is obtained using large electrodes with large inter-electrode distance, the minimum of ARV of the signals detected over the IZ is less marked and may become a ripple. Even in this case estimates of MNF are sensitive to electrode position and estimates of average CV may be impossible. The scatter of IZs may be different in different muscles and, for the same muscle, in different subjects. This is the reason why a minimum of ARV or RMS has been detected by some authors and not by others. In any case, the general rule of avoiding the IZ for placement of an electrode pair should be followed. Of course, as indicated by Campanini et al. (2007), the criticality of the issue depends on the application but, nevertheless, it is a good practice to follow the general rule. In dynamic conditions the IZ moves under the skin and if it reaches the electrodes it may cause changes of EMG amplitude that are easily misinterpreted as changes of muscle activation level (Kiryu et al., 1996; Martin and MacIsaac, 2006; Schulte et al., 2004). In dynamic conditions it is therefore important to verify that the IZ does not approach the detection electrodes during the entire range of movement.

Some authors did not recognize this problem (Beck et al., 2009) since large electrodes were adopted in their work. Such electrodes introduce low pass filtering and spatial averaging resulting in limitations in the quality of the estimated EMG variables. For these reasons we strongly suggest to avoid electrode placing over the IZ and rather locating them in between the IZ and tendon zones.

The individual variations of limb and muscle anatomy require efforts to minimize the resulting variability of EMG features. It is foreseen that surface EMG computer assisted systems may soon detect EMG from two dimensional arrays (High Density EMG) and automatically identify the electrode pairs or groups that are most suitable as a source of information concerning the location of IZs, the anatomy of individual active MUs, the fiber direction and the features of surface EMG that are relevant for meaningful physio-pathological observations and conclusions.

The growing use of electrode arrays creates a need for both continuing education of clinical users and for the definition of standards. The need for a standardization effort, continuing the endeavors of Project SENIAM, is perceived worldwide. A recent meeting held at the Worcester Polytechnic Institute (Massachussetts, USA) concerning multichannel detection, decomposition and interpretation of EMG constitutes a first effort in this direction [www.emglab.stanford.edu] that should be supported and expanded.

In conclusion, our results show that when a single electrode pair is used the IED must be small with respect to the distance between the IZ and the tendon and neither electrode of the pair should be over the IZ for the entire range of the movement. If this is not the case, EMG variables may be highly affected by small geometrical changes.

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Luca Mesin graduated in Electronics Engineering in December 1999 from Politecnico di Torino, Torino, Italy, he received the Ph.D. in Applied Mathematics in 2003, from the same University. Since June 2003, he is a research assistant of the Laboratory for Neuromuscular System Engineering in Torino. He was involved in research activities in the fields of Kinetic theory and Deformable Porous Media theory, with applications to Biomathematics and Composite Materials. Now, his main research interests concern signal processing of biomedical signals and modeling of biological systems.

Roberto Merletti graduated in Electronics Engineering from Politecnico di Torino, Italy and obtained his M.Sc. and Ph.D. from The Ohio State University. He is Professor of Biomedical Instrumentation at the Dept. of Electronics of Politecnico di Torino and Director of the Laboratory for Neuromuscular System Engineering in Torino, Italy. He has been Associate Professor at the Department of Biomedical Engineering and at the Neuro Muscular Research Center of Boston University. His research focuses on surface electromyography, myoelectric signal processing and electrical stimulation.

Alberto Rainoldi graduated in Physics (1991) from Torino University. He obtained the Ph.D. in Physical Medicine and Rehabilitation (2002) at the Faculty of Medicine, Università di Roma "Tor Vergata", Italy. Since 1996 he has been with Politecnico di Torino, where his research activity is focused on: the assessment of proper methodologies useful to increase the standardization and the repeatability of surface EMG measures, allowing their clinical use. He is also a scientific consultant of the S. Maugeri Foundation (Pavia, Italy) for the research projects in the field of Muscle Fatigue. Since 2004 he is professor at the Motor Science School of the University of Turin, Italy where he teaches the course of "Research Methodology" and "Bioengineering". Since 2007 he leads the Motor Science Research Center in that Faculty. He is member of the Editorial Board of the Journal of Electromyography.