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1 **Title:** Insights gained into the interpretation of surface electromyograms from the gastrocnemius
2 muscles: a simulation study.

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1 **ABSTRACT**

2 Interpretation of surface electromyograms (EMG) is usually based on the assumption that the
3 surface representation of action potentials does not change during their propagation. This
4 assumption does not hold for muscles whose fibers are oblique to the skin. Consequently, the
5 interpretation of surface EMGs recorded from pinnate muscles unlikely prompts from current
6 knowledge. Here we present a complete analytical model that supports the interpretation of
7 experimental EMGs detected from muscles with oblique architecture. EMGs were recorded from
8 the medial gastrocnemius muscle during voluntary and electrically elicited contractions.
9 Preliminary indications obtained from simulated and experimental signals concern the spatial
10 localization of surface potentials and the myoelectric fatigue. Specifically, the spatial distribution of
11 surface EMGs was localized about the fibers superficial extremity. Strikingly, this localization
12 increased with the pinnation angle, both for the simulated EMGs and the recorded M-waves.
13 Moreover, the average rectified value (ARV) and the mean frequency (MNF) of interference EMGs
14 increased and decreased with fatigue, respectively. Furthermore, the degree of variation in ARV and
15 MNF did not depend on the pinnation angle simulated. Similar variations were observed for the
16 experimental EMGs, although being less evident for a higher fiber inclination. These results are
17 discussed on a physiological context, highlighting the relevance of the model proposed here for the
18 interpretation of gastrocnemius EMGs and for conceiving future experiments on muscles with
19 pinnate geometry.

20

1 **1. INTRODUCTION**

2 The modeling of surface electromyograms (EMGs) has been sought for the interpretation of
3 experimental data (Dimitrova and Dimitrov 2003; Merletti et al., 1999b; Roeleveld et al., 1997), for
4 the development of algorithms aimed at information extraction (Duchene and Hogrel, 2000; Mesin
5 et al., 2009), and for didactic purposes (Merletti et al., 1999a). Available models for the generation
6 of surface EMGs rely both on numerical (Lowery et al., 2004; Mesin et al., 2006) and analytical
7 (Block et al., 2002; Farina et al., 2004b; Gootzen et al., 1991; Mesin, 2006) approaches.

8
9 The assumption that the volume conductor is invariant in the direction of propagation of
10 intracellular action potentials allowed for the development of fast analytical models for the
11 simulation of surface EMGs (Farina and Merletti, 2001; Farina et al., 2004a). If a volume conductor
12 is space invariant, the surface representation of a motor unit action potential does not change with
13 its propagation. Such space invariance is frequently assumed when simulating, processing and
14 interpreting experimental EMGs (Lindstrom and Magnusson, 1977; Reucher et al., 1987).
15 Therefore, much of the insights gained into the interpretations of surface EMGs from the use of
16 mathematical models are valid, exclusively, for muscles whose geometry fits in the assumption of
17 space invariance.

18
19 Some of the muscles investigated by surface EMG cannot be approximated assuming the muscle
20 fibers to be parallel to the skin. The fibers of the gastrocnemius muscle, for example, are inclined
21 with respect to the skin surface and extend from the deep to the superficial aponeurosis (Kawakami
22 et al, 1998; Narici et al., 1996). Surface electrodes positioned on the calf are, thus, located above the
23 superficial aponeurosis, where muscle fibers attach. Considering that action potentials propagate
24 along the muscle fibers, their propagation along the oblique gastrocnemius fibers contributes to the
25 surface EMGs with a component toward muscle extremities and another toward the skin/tibia.
26 Therefore, the area on the skin surface upon which the action potential of a single muscle fiber

1 distributes likely depends on how inclined the muscle fibers are (i.e. its pinnation angle) (Vieira et
2 al., 2011). Currently existing models do not provide indications of how the surface EMG relates to
3 the pinnation angle. Which physiologically relevant information might be extracted from surface
4 EMGs in the pinnate gastrocnemius muscle is unknown.

5
6 In this study we simulate a pinnate muscle, with fibers inclined in the depth direction, to interpret
7 surface EMGs detected from the gastrocnemius muscles. A complete mathematical model, which
8 includes muscle fibers with finite length and three layers of tissues, is presented here for the
9 generation of single fiber action potentials (SFAP). In particular, we simulate the distribution of
10 motor unit action potentials (MUAPs) on the skin and the interference EMG for different degrees of
11 inclination of muscle fibers. The implications for the interpretation of how action potentials
12 distribute on the skin surface and for the estimation of muscle fatigue are addressed as well.
13 Ultrasound images and experimental EMGs are recorded from the human medial gastrocnemius
14 (MG) muscle to investigate how much theoretical and empirical data correspond.

15

16 **2. METHODS**

17 **2.1 Mathematical model**

18 The simulation model was developed by extending our previous work (Mesin and Farina, 2004),
19 which considered the impulse response of a two layer volume conductor. Three layers were
20 considered here (skin, fat and muscle; figure 1). Moreover a complete model (including finite-
21 length fibers) was implemented to generate single fiber action potentials (SFAP).

22

Figure 1

23 Libraries of MUAPs were generated from the simulated SFAPs. These libraries were used with a
24 model of spatial and temporal recruitment of motor units (MUs) to simulate interference EMGs
25 during fatiguing contractions.

26

1 Full details on the model and the simulated EMGs are described in the Appendix.

2

3 **2.2 Experimental signals**

4 Single-differential EMGs were recorded from the medial gastrocnemius (MG) muscle to test for the
5 correctness of the information gained from simulated signals. In particular, we investigated how the
6 amplitude of surface potentials varies with the pinnation angle. This is of remarkable interest since
7 this variation likely provides information of how localized the gastrocnemius activity might be.
8 Moreover, we were interested in understanding whether experimental and simulated EMGs in
9 pinnate muscles are comparable during fatiguing contractions.

10

11 Three male subjects (age: 33, 29, 27years; body mass: 78, 80, 75kg; height: 182, 178, 180cm)
12 participated in two protocols designed to compare experimental and simulated signals.

13

14 Protocol 1: Sixteen surface electrodes (10 mm IED) were used to record EMGs during electrical
15 stimulation. Two ankle angles were considered; foot in neutral position ($\sim 20^\circ$ pinnation angle) and
16 plantar flexed ($\sim 35^\circ$ pinnation angle). With an adhesive pre-gelled electrode (cathode; figure 6a)
17 placed carefully on the leg, bipolar current pulses were delivered at 2 pps and for 20s to the tibial
18 posterior nerve. The anode electrode (80×50 mm; soaked cloth) was positioned immediately above
19 the patella with elastic Velcro straps. Stimulation amplitude was as minimal as possible to allow for
20 the detection of the firstly observable M-wave. Low stimulation amplitude was chosen to recruit the
21 least number of MUs and, thus, to better isolate the effect of ankle angle on the distribution of M-
22 waves amplitude on the skin. Averaged M-waves were obtained after triggering the 15 single-
23 differential EMGs.

24

25 Protocol 2: EMGs were recorded with an array of eight electrodes (5mm interelectrode distance –
26 IED) when the subjects exerted isometric plantar flexion at 60% MVC. The contraction lasted for

1 30s to ensure the occurrence of myoelectric manifestation of fatigue. Fatigue plots were created
2 from the average rectified value (ARV) and the mean frequency (MNF) calculated on 500ms
3 epochs (i.e. both descriptors are plotted with respect to their initial values). ARV and MNF indices
4 were calculated as indicated in Merletti et al. 1990. This protocol was applied once with the foot
5 dorsal flexed (pinnation angle $\sim 10^\circ$) and once in neutral position (pinnation angle $\sim 20^\circ$), with 5 min
6 interval between trials. Plantar flexion torque was measured and displayed to the subjects. Subjects
7 were in prone position and their feet were firmly secured to a footplate, with the lateral malleolus
8 being coaxial to the centre of rotation of the torque meter. MVC values were determined for each
9 pinnation angle as the maximum torque measured across three attempts separated by 5 minutes.

10

11 EMGs were amplified (gain ranged from 1k to 5k; 10–500 Hz EMG-USB amplifier, LISiN and
12 OTBioelettronica, Turin), and sampled at 2048 Hz with a 12bit A/D converter ($\pm 2.5V$ dynamic
13 range). Ultrasound images were taken with a linear probe (3.86 cm long; Fukuda Denshi, UF 4000,
14 7.5 MHz) and pinnation angles were estimated with the precision of one degree. A custom-made
15 neuromuscular stimulator (LISiN, Turin), equipped with a hybrid output stage, was used in the first
16 protocol. In the second protocol, load cells output was amplified (150Nm full-scale amplifier;
17 MISOI, OTBioelettronica, Turin) and then converted to values of ankle torque. After cleansing of
18 the skin with abrasive paste and water, ultrasound scanning was used to place electrodes on the MG
19 muscle. Specifically, electrodes were positioned above the sheath of aponeurotic tissue and parallel
20 to the surface projection of the pinnate MG fascicles. Care was taken to ensure that the array and
21 the ultrasound probe had similar orientation, which provided an ultrasound image with as many
22 fascicles as possible.

23

24 **3. RESULTS**

25 **3.1 Preliminary Simulations**

1 Some preliminary simulations are shown in figures 2 and 3, in order to compare the surface
2 potentials produced by fibers with different arrangements (parallel or slightly inclined with respect
3 to the skin). In particular, the potential distribution over the skin surface is considered for fixed time
4 samples in figure 2, whereas the signals detected as functions of time from arrays of electrodes
5 sampling the potential in specific points aligned to the fibers are shown in figure 3.

6
7 Simulations showed that the amplitude and the shape of surface potentials change with the obliquity
8 of muscle fibers (figure 2). When simulating fibers parallel to the skin, the surface distribution of
9 action potentials was symmetric with respect to the innervation zone (IZ). Conversely, by
10 increasing the fiber inclination, the surface potential became more asymmetric, as the contribution
11 of the deeper source was attenuated and more diffused than that of the most superficial source. The
12 root mean square (RMS) difference between the surface distribution of potentials simulated for 10°
13 and those computed for the other pinnation angles (0° , 5° , 15° , and 20°) is shown as a function of
14 the source position (figure 2b). As expected, greater RMS differences were obtained for higher
15 variations between angles. The greatest difference occurred when the source was close to the
16 extinction region.

17
18 Regardless of whether simulating monopolar or single-differential signals, variations in the
19 amplitude of surface potentials across electrodes depended on the pinnation angle. By simulating an
20 electrode array placed parallel to the x-z projection of a superficial muscle fiber with 0° pinnation
21 angle, both monopolar and differential potentials appeared with equal amplitudes on either sides
22 from the IZ (left panel in figure 3a). For deep fibers, only the standing waves produced by the
23 generation and extinction of simulated potentials are seen in the EMGs, respectively (right panel in
24 figure 3a). By inclining muscle fibers, the amplitude distribution of surface potentials concentrated
25 progressively more toward the superficial tendon (left panels in figure 3b,c). Interestingly, pinnate
26 fibers simulated 10cm away from the electrodes contributed to the surface EMGs with potentials of

1 markedly smaller amplitude (3-5 times smaller; right panels in figure 3b,c). Therefore, surface
2 electrodes on pinnate muscles sample only from sources located nearby and nowhere else.

3
4 The inclination of muscle fibers affects also the shape and the propagation of surface potentials.
5 Close inspection of figure 3 reveals that the surface potential from fibers parallel to the skin has the
6 same duration, wherever it is sampled on the skin. Propagation of these potentials is also evident in
7 the figure. Potentials from pinnate fibers undergo large shape variations when recorded from
8 different channels, so that conduction velocity (CV) cannot be estimated properly (Farina and
9 Merletti, 2004). Specifically, they have shorter duration when detected closer to the superficial
10 tendon. Moreover, the delay between successive peaks or valleys of potentials simulated for
11 adjacent channels is not the same for pinnate fibers. Then, surface EMGs detected from muscles
12 with fibers not parallel to the skin unlikely show the propagation of action potentials along muscle
13 fibers.

14 **Figures 2 and 3**

15

16 **3.2 Amplitude distribution of EMGs**

17 In figure 4, simulated single fiber potentials are investigated to quantify how much the spatial
18 distribution of surface EMGs changes as a function of the pinnation angle. Surface potentials were
19 compared for volume conductors with various pinnation angles, from 0° to 45°, and for two
20 thicknesses of the fat layer. Contour plots of simulated surface potentials are shown at a given time
21 (figure 4b). To quantify the diffusion of the surface potential, the median and the first and third
22 quartiles were investigated along orthogonal sections of contour plots (figure 4c). The potential
23 generated using a model of parallel fibers was more diffused (interquartile interval: ~40mm) over
24 the skin with respect to that generated with a pinnate geometry (interquartile interval: less than
25 25mm for pinnation angles higher than 10°; figure 4).

26

1 As for the simulated SFAPs, the amplitude distribution of electrically elicited EMGs depended on
2 the degree of inclination of MG fibers. When MG fascicles were $\sim 20^\circ$ oblique, the firstly emerging
3 M-waves appeared on the most proximal muscle portion, from the channel 1 to 5 (figure 5b). The
4 ARV amplitude of M-waves distributed evenly across these channels (interquartile interval:
5 53.7mm). Strikingly, for the same stimulation amplitude and higher pinnation angle ($\sim 35^\circ$), the
6 distribution of ARV amplitude changed chiefly across channels (interquartile interval: 32.2mm),
7 with the most proximal channels detecting the largest M-waves (figure 5c). Additionally, M-waves
8 recorded from the pinnate gastrocnemius muscle showed neither a delay nor a phase opposition
9 between consecutive channels.

10 **Figure 4 and 5**

11 **3.3. Simulated and experimental manifestation of fatigue in surface EMGs**

12 Simulations indicate that amplitude and spectral descriptors might be used to study the myoelectric
13 manifestations of fatigue in the pinnate gastrocnemius muscle. Single-differential EMGs, simulated
14 and recorded with seven channels aligned to the longitudinal projection of muscle fibers, are shown
15 for two pinnation angles (10° and 20° ; figure 6). Both simulated (session 2.4) and experimental
16 signals showed myoelectric manifestation of fatigue, with ARV and MNF increasing and
17 decreasing with time, respectively (fatigue plots; figure 6). Interestingly, changes in amplitude and
18 frequency were more variable for the experimental EMGs; and this high variability increases with
19 the pinnation angle (Table 1).

20 **Figure 6, Table 1**

21 **4. DISCUSSION**

22 A model for the simulation and interpretation of surface EMGs in pinnate muscles is proposed here.
23 Simulations indicate that the obliquity of muscle fibers has a marked effect on the surface EMG
24 (figure 2). The higher the fibers pinnation the more localized was the surface distribution of action
25 potentials, with electrodes closer to the fibers end detecting remarkably higher potentials. This
26 localized distribution of surface potentials is more evident for single-differential than for monopolar

1 signals, likely due to the higher selectivity of differential derivations. Experimental EMGs from the
2 MG substantiated the predictions posed by our simulations.

3
4 While there are architectural differences between skeletal muscles, it seems worthy to ask whether
5 the surface EMGs recorded from pinnate muscles are as informative as those recorded from
6 fusiform muscles. Understanding how the amplitude distribution of surface potentials changes with
7 the pinnation angle and the possibility of using surface EMGs to study muscle fatigue are of
8 particular interest.

9
10 **4.1 Localization of surface EMGs depends on the pinnation angle of the gastrocnemius muscle**

11 Our simulations revealed that surface electrodes on muscles with fibers not parallel to the skin
12 sample from nearby sources and from nowhere else. This key result is substantiated from our
13 observations that: i) surface potentials were localized over the superficial tendon of pinnate fibers,
14 with the main contribution being due to the extinction phase of intracellular action potentials (figure
15 3); ii) although the representation of surface potentials was 50% less localized when trebling the fat
16 thickness, it was still considerably more concentrated on the superficial tendon than the surface
17 representation of potentials in parallel fibers (figure 4); iii) two pinnate fibers with distal tendons
18 separated by 10cm produced surface potentials with separated amplitude distributions.
19 Consequently, the pinnation brings a wider cross section of the muscle into the view of surface
20 electrodes than that available for parallel-fibered muscles. It is worth to mention that the localized
21 distribution shown in figures 3 and 4 refers to the amplitude of SFAPs. As the amplitude of MUAPs
22 corresponds to the algebraic summation of several SFAPs, the localized representation of individual
23 MUAPs in the surface EMGs depends on whether MUs have small territories (Vieira et al., 2011),
24 Here, we are not interested on the size of MUs territories but on how much the amplitude
25 distribution of SFAPs changes with the pinnation angle.

26

1 Electrical stimulation of the tibial nerve revealed a somewhat similar localization of surface M-
2 waves detected from the MG muscle (figure 5), with respect to that observed for simulated
3 potentials. The smallest stimulation amplitude leading to the first observable M-waves was chosen
4 to ensure that only a few MUs were stimulated. By keeping constant the stimulation amplitude, the
5 effect of pinnation angle on the localization of M-waves was isolated. M-waves with similar
6 amplitude were observed for the four most proximal channels when stimulation pulses were
7 delivered with the foot in neutral position (figure 5b, $\theta = 20^\circ$). With the foot plantar flexed (figure
8 5c, $\theta = 35^\circ$), the amplitude of M-waves distributed unevenly across the four most proximal
9 channels. Larger M-waves appeared for the more proximal channels. While the variation in ankle
10 joint angle could have induced variation in the stimulation site, as the nerve moves beneath the
11 stimulation electrode, the similitude of EMGs likely indicates that the same population of MUs was
12 stimulated in both foot conditions.

13

14 Surface EMGs convey unique information regarding the activation of the pinnate gastrocnemius
15 muscle. In muscles whose fibers are parallel to the skin, surface electrodes detect the propagation of
16 MUAPs (Farina et al., 2002; Merletti et al., 2003). In the gastrocnemius muscle, the amplitude of
17 surface EMGs varies with the number of active fibers beneath the recording electrodes. Very
18 recently, for example, we used the model presented here to validate our estimations of the
19 longitudinal size of MUs territory in the human MG muscle (Vieira et al., 2011). Specifically, we
20 observed that with respect to the MG length, the fibers of individual MUs extended for only a short
21 distance (less than 4cm). Then, activation of individual MUs leads to regional activation of the MG
22 muscle. Indeed, the extensive evidence positing localized activation of the calf muscles in humans
23 and cats is not surprising (Eng and Hoffer, 1997; English and Weeks, 1989; McLean and Goudy,
24 2004; Staudenmann et al., 2009; Vieira et al., 2010a,b; Wolf et al., 1998). If the nervous system
25 takes advantage of muscle architecture to shape the recruitment of MUs (Kennedy and Cresswell,

1 2001; Vieira et al., 2011), the high-density surface electromyography (Merletti et al., 2010) could
2 provide a mean to study the regional organization of activity in the pinnate gastrocnemius muscle.

3

4 **4.2 Myoelectric manifestation of fatigue in pinnate muscles**

5 Testing for the possibility of studying fatigue manifestation in skeletal muscles from the surface
6 EMG is not simple. Several factors affect the muscles undergoing fatigue: 1) different strategies of
7 MU recruitment and firing rate; 2) variations in load sharing between synergistic muscles,
8 especially for those spanning the same joint as soleus and gastrocnemius (McLean and Goudy,
9 2004); 3) changes in shape of intracellular action potentials (Arabadzhev et al., 2005); 4)
10 synchronization of MUs (Mesin et al., 2009); 5) variable decrease of CV for muscle fibers of
11 different physiological types (Rainoldi et al., 2008). To keep simulations simple, we considered the
12 same percentage of CV changes for all MUs. Additionally, the same spatial and temporal
13 recruitment of MUs was simulated for two different pinnation angles, so as to isolate the effect of
14 fibers inclination on surface EMGs.

15

16 Even though it was not possible to observe the changes in CV directly, it was possible to observe
17 myoelectric manifestations of gastrocnemius fatigue in the amplitude and spectral EMG descriptors.
18 Because of the muscle oblique architecture, and considering the scattering of IZs, the extinction of
19 action potentials at the superficial aponeurosis contributed chiefly to the surface EMGs simulated
20 (figure 3). As a consequence, electrodes located over the aponeurotic layer do not detect the same
21 potential propagating toward either the superficial or deep aponeurosis. Instead, they record
22 standing waves likely resulting from the end-of-fiber effect (Stegeman et al., 1997). This
23 interpretation is supported by spike-triggered EMGs, which did not show physiological delays
24 between surface potentials detected by consecutive electrodes located upon the superficial
25 aponeurosis (Vieira et al., 2011). Thus, estimation of CV, which showed remarkable properties in
26 the investigation of myoelectric fatigue in parallel-fibered muscles (Mesin et al., 2009), cannot be

1 obtained from the gastrocnemius muscle, at least not upon its superficial aponeurosis. Nevertheless,
2 simulations showed that ARV and MNF of surface EMGs are both affected by variations in CV.

3
4 Although at different extents, the experimental and simulated manifestations of fatigue were
5 observed in the pinnate gastrocnemius muscle. Simulations did not show any reliance of the
6 variations in ARV and MNF on the pinnation angle. Nevertheless, the variability of ARV and MNF
7 were different for experimental data recorded for two pinnation angles (Table 1). This suggests that
8 some factors, which were kept constant in the simulations, were involved in the contraction of the
9 MG muscle at different pinnation angles. It has been shown, for example, that variations in activity
10 within and between calf muscles occur when subjects are asked to sustain a low and isometric
11 torque of plantar flexion (McLean and Goudy, 2004; Tamaki et al., 1998). Variable activation
12 within the same gastrocnemius muscle was observed even for different directions of ankle force
13 (Staudenmann et al., 2009) and in quiet standing (Vieira et al., 2010a,b). Considering the more
14 localized distribution of surface potentials for the more oblique geometry (figures 4), and the small
15 territories of gastrocnemius MUs (Vieira et al., 2011), surface electrodes seem to provide very
16 selective recordings for high pinnation angles. Therefore, variations in recruitment and firing rate of
17 MUs during a fatiguing contraction would affect more severely the surface EMGs detected from
18 more oblique MG fibers.

19

20 **4.3 Future perspectives**

21 The model presented here opens new perspectives for investigating the calf muscles activation,
22 either by supporting the interpretation of experimental data or by providing theoretical grounds
23 upon which future studies will be designed. The surface distribution of simulated action potentials
24 (figures 3 and 4), for example, supports our prediction that postural activation of the MG muscle is
25 organized regionally for the control of quiet standing posture (Vieira et al., 2010a). Similarly, our

1 simulation results indicate that amplitude and spectral EMG descriptors could be potentially useful
2 to study fatigue in the gastrocnemius muscle.

3

4 **5. CONCLUSIONS**

5 A new model for the simulation of surface EMGs in pinnate muscles was developed and applied to
6 interpret experimental data obtained from the gastrocnemius muscles. The most striking result was
7 the localized representation of surface potentials. Because of the gastrocnemius pinnate
8 architecture, simulated EMGs and experimental M-waves were both detected only by few
9 consecutive electrodes. Higher pinnation angles led to more localized potentials. The regional
10 organization of gastrocnemius activity in different motor tasks could, then, be investigated with
11 surface electromyography. The potentiality of the model was also shown for the investigation of
12 myoelectric fatigue with amplitude and spectral indexes. Both descriptors varied during fatigue,
13 either for experimental or simulated EMGs. The greater variability in the amplitude and frequency
14 of experimental EMGs, observed for higher pinnation angles, was likely due to different
15 recruitment strategies.

16

17 **Conflict of Interest Statement**

18 None of the authors has any conflict of interest concerning the publications of this work.

19

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24

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1 **Table 1:** Variability of the amplitude (ARV) and frequency (MNF) descriptors of experimental and
2 simulated surface EMGs, calculated as the standard deviation of residuals resulting from the fitting of
3 a first order polynomial to the observed data (dashed lines in figure 4).

	Experimental EMGs		Simulated EMGs	
Pinnation angle (deg)	ARV (μV)	MNF (Hz)	ARV (a.u.)	MNF (Hz)
10	67	4.16	4.21	3.61
20	108	4.68	4.20	3.62

4

5

1 **FIGURE CAPTIONS**

2 **Figure 1** a) Schematic representation of the volume conductor model, including the definition of
3 the coordinate system and the geometry of the simulated muscle fibers. The muscle layer is
4 considered as homogeneous and anisotropic, with fibers inclined with respect to the skin surface.
5 Fat and skin layers are homogeneous and isotropic. b) Ultrasound image of the gastrocnemius,
6 showing the pinnation of muscle fibers. c) Longitudinal section of the simulated volume conductor,
7 indicating the notation used for determining the analytical solution. d) Sampling of the
8 phenomenological model of the transmembrane current proposed in Rosenfalck (1969), using 10
9 impulse sources.

10 **Figure 2** a) Surface potential generated in muscle fibers with three different pinnation angles (0° ,
11 5° , and 10°). The signals were simulated using the model shown in Figure 1a,c. b) Root mean
12 square (RMS) difference between the EMG map generated for the pinnation angle of 10° and the
13 maps computed for pinnation angles of 0° , 5° , 15° , and 20° . Two different thicknesses of the fat
14 layer were simulated. RMS values are expressed in percentage and as a function of the position of
15 the source (measured as the distance of each of the two transmembrane current sources from the
16 innervation zone).

17 **Figure 3** Simulation of surface EMG signals detected with an electrode array located over the
18 longitudinal projection of a single muscle fiber on the skin surface. Monopolar and single-
19 differential signals are shown for different fibers, corresponding to different pinnation angles or to
20 different positions.

21 **Figure 4** Simulated (a; see Appendix) and experimental (b; see section 2.2) EMGs are shown for
22 two pinnation angles (10° and 20°) during an isometric fatiguing contraction at 60% of maximal
23 voluntary contraction (MVC). Interference EMGs were simulated with the muscle fiber conduction
24 velocity - CV - decreasing by 1%/s of the initial value. The volume conductors simulated and
25 ultrasound images of the MG muscle are shown on top (arrows indicate detection points). The same
26 firing pattern and population of 340 motor units were simulated for two pinnation angles (10° and

1 20°). A short time epoch of raw signals is depicted in the middle panel. Fatigue plots are shown in
2 the bottom, including averaged rectified values (ARV) and mean frequency (MNF), estimated for
3 epochs of 500 ms duration (n = 60 epochs) and for each pinnation angle. Values were averaged
4 across channels (n = 7 channels). Dashed lines indicate the best linear fitting to ARV and MNF
5 traces, calculated with the least square method.

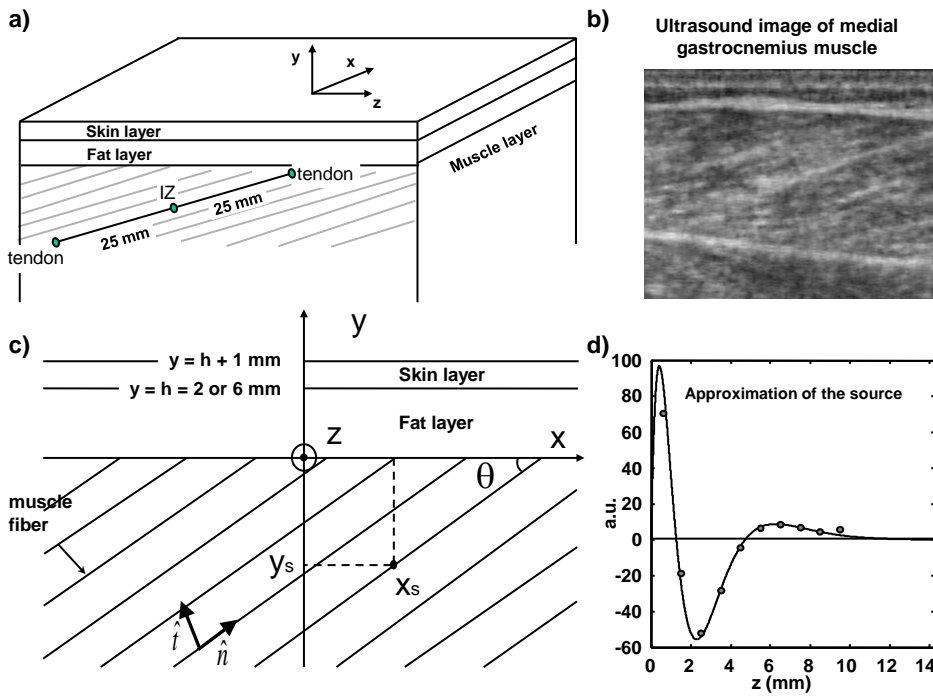
6 **Figure 5** Spatial distributions of surface potentials as a function of the pinnation angle. a)
7 Longitudinal view of the volume conductors considered. When the pinnation angle is higher than
8 zero, one of the tendons is located at the muscle/fat interface, representing the superficial
9 aponeurosis. The case in which fibers are parallel to the skin surface is also considered, for two
10 muscle depths. b) Contour plots of the rectified value, averaged over time (ARV), of the simulated
11 surface potential, indicating the median and the first and third quartiles for orthogonal sections
12 crossing the location of maximal amplitude. Note the asymmetry of contours for pinnation angles
13 greater than zero; ARV amplitude concentrates close to the fiber end (the superficial aponeurosis).
14 c) Median, first and third quartiles of ARV in the direction longitudinal (x) and transversal (z) to the
15 fibers are shown for different pinnation angles and for two fat layers.

16 **Figure 6** Spatial distribution of single-differential M-waves detected from the medial
17 gastrocnemius (MG) muscle at two pinnation angles. a) Array of 16 surface electrodes, positioned
18 on the MG muscle, and the stimulation electrode, located above the tibial nerve branch supplying
19 the same muscle. Note that the most proximal EMG electrodes were nearer to the stimulation
20 electrode. b) Raw triggered (black traces) and averaged (gray traces) M-waves detected with the
21 foot positioned so as to result in a pinnation angle of about 20°. The ultrasound image obtained with
22 the foot in this same position is shown on the top. c) the same as in b) for ~35° pinnation angle.
23 Note that the stimulation artefact was most evident for the most proximal channels, which were the
24 closest to the cathode electrode.

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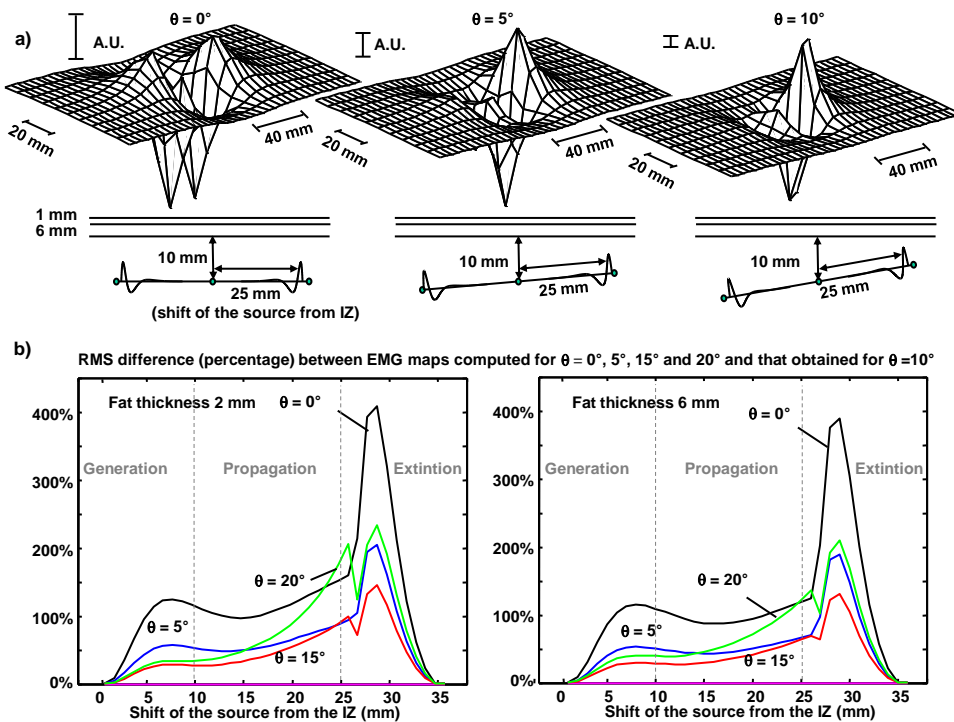
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1 Fig 1



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3 Fig 2



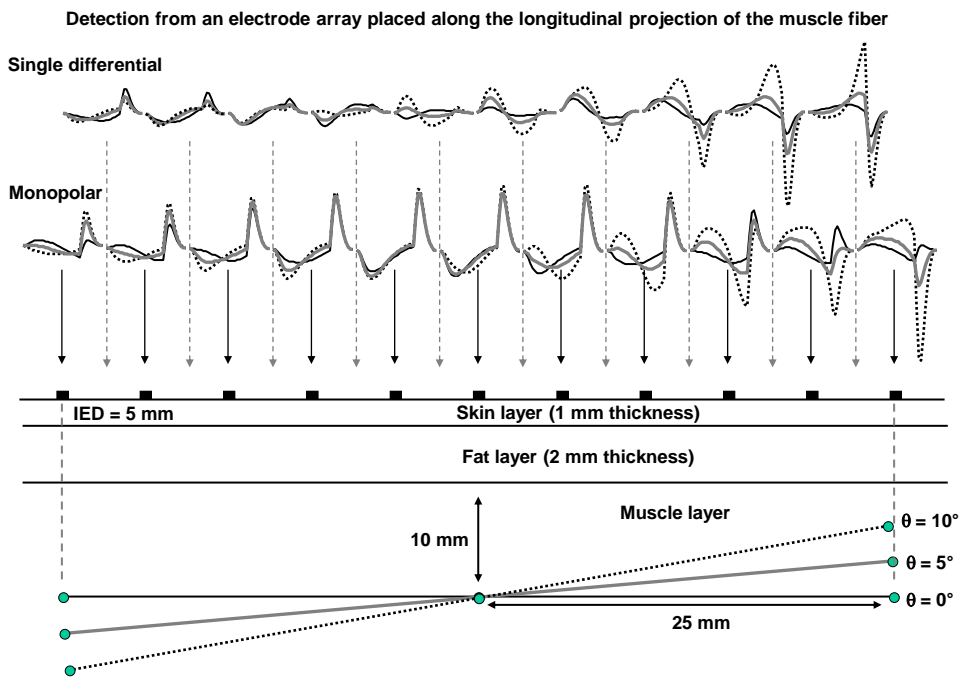
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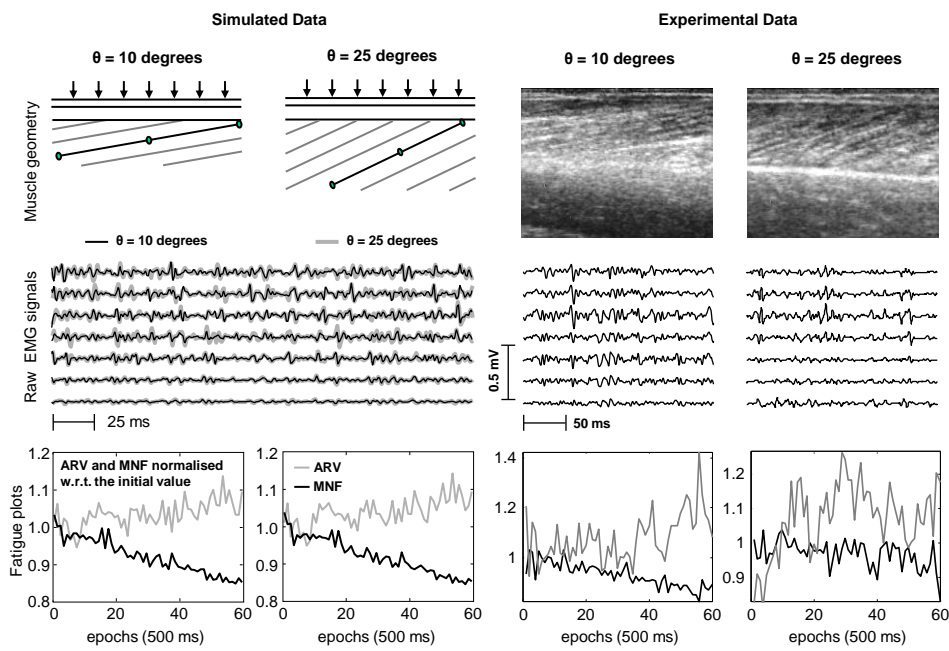
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1 Fig 3



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3 Fig 4



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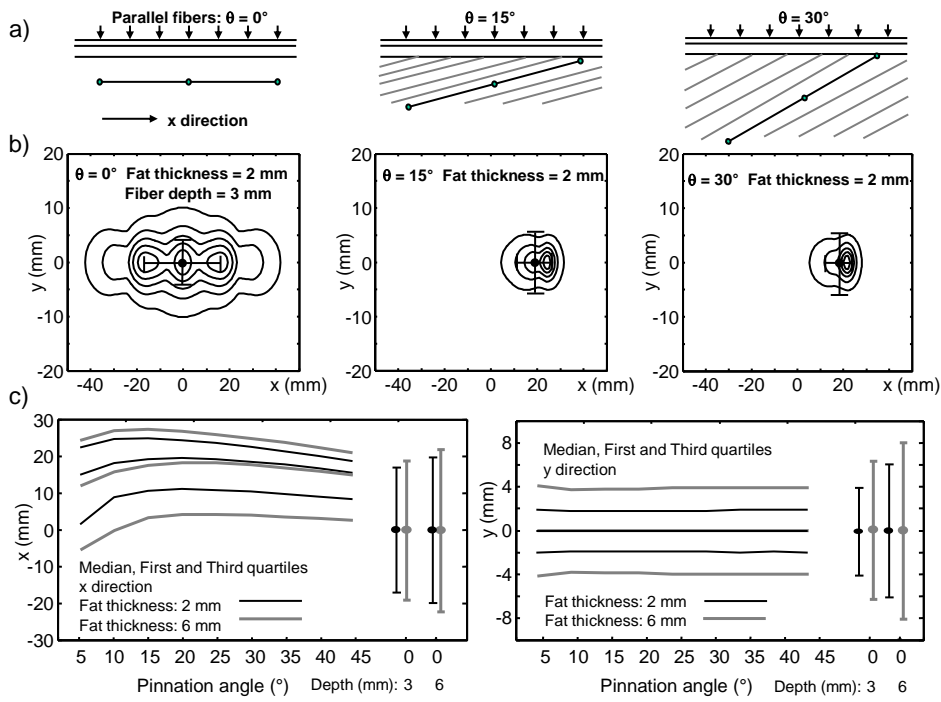
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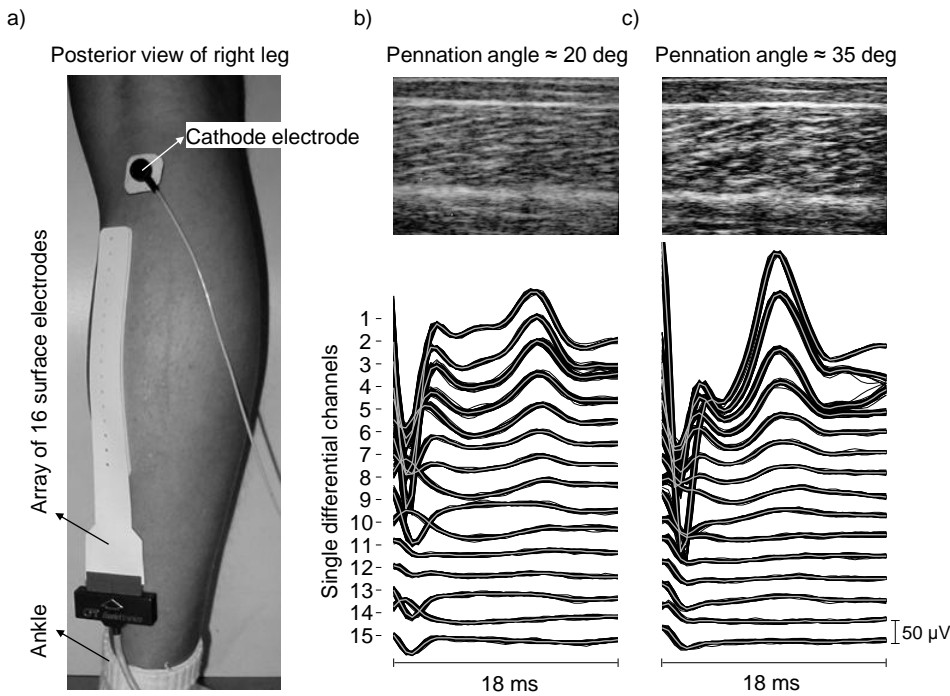
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2 Fig 5



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4 Fig 6



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