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MODEL OF GENERATION OF SURFACE EMG WITH MULTI-LAYER VOLUME CONDUCTOR WITH VARIABLE THICKNESS OF SUBCUTANEOUS TISSUE

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1. INTRODUCTION

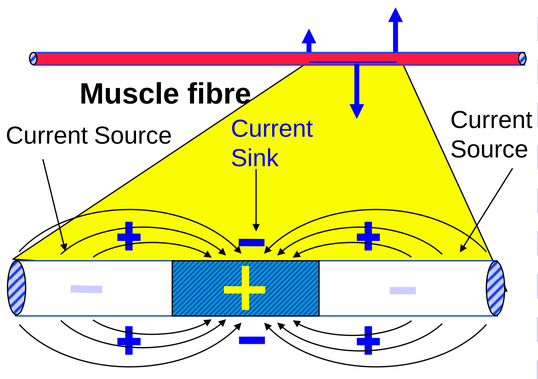
Simulation of surface electromyography (EMG) found application in the estimation of physiological variables (inverse problem), the choice of the detection system, the design and test of algorithms for information extraction, the interpretation of experimental results. Both analytical and numerical methods have been proposed.

A structured generation model for EMG requires to address the following issues

A) Analysis of the Source

Transmembrane current

Equivalent tripole



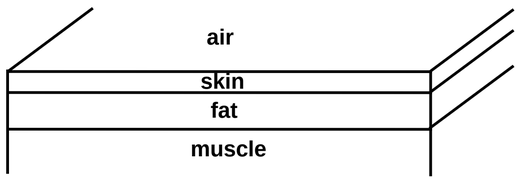
B) Mathematical Problem: conductivity tensor for EMG, influence of geometry, boundary conditions

Poisson problem. Homogeneous Neumann's conditions at the boundaries

$$\nabla \cdot J = -\nabla \cdot (\underline{\underline{\sigma}} \nabla \varphi) = I \qquad \hat{n} \cdot \nabla \varphi = 0$$

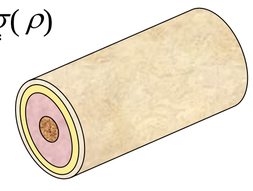
Plane layers [1]

$$\underline{\underline{\sigma}} = \sigma_{xx}\vec{i} + \sigma_{yy}\vec{j} + \sigma_{zz}\vec{k}$$



Cylindrical layers [2]

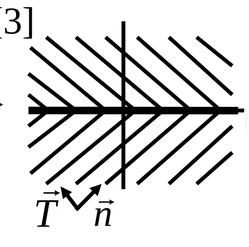
$$\underline{\underline{\sigma}}(\rho,\theta,z) = \underline{\underline{\sigma}}(\rho) \qquad \underline{\underline{\sigma}} = \sigma_{\rho\rho}\vec{i} + \sigma_{\theta\theta}\vec{j} + \sigma_{zz}\vec{k}$$



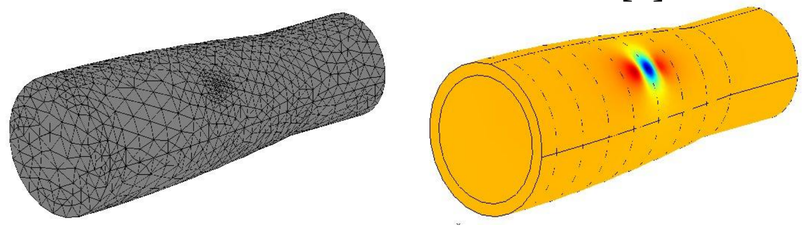
Pinnate Muscle [3]

$$\underline{\underline{\sigma}} = \sigma_{nn}\vec{n} + \sigma_{TT}\vec{T} + \sigma_{zz}\vec{k}$$

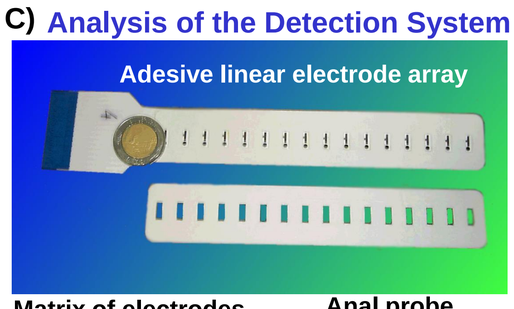
$$\underline{\underline{\sigma}}(x,y,z) = \underline{\underline{\sigma}}(z)$$



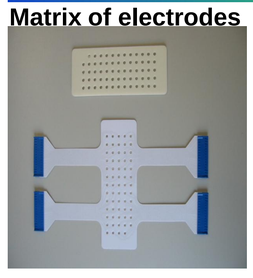
Fusiform Muscle with curvilinear fibres [4]



C) Analysis of the Detection System



Matrix of electrodes



Anal probe

Single Differential: +1 -1

Double Differential: -1 +2 -1

Laplacian: -1 / -1 +4 -1 / -1

SD DD NDD

Fig.1 Basic steps in the development of a structure based model of generation of surface EMG signals. A) Source of the problem. B) Poisson problem for different volume conductors. C) Detection system.

2. METHODS

A two layer volume conductor describing subcutaneous tissue and muscle tissues is considered. The subcutaneous tissue is assumed isotropic, the muscle anisotropic. Muscle layer is a hemi-space. Subcutaneous tissue has a variable thickness.

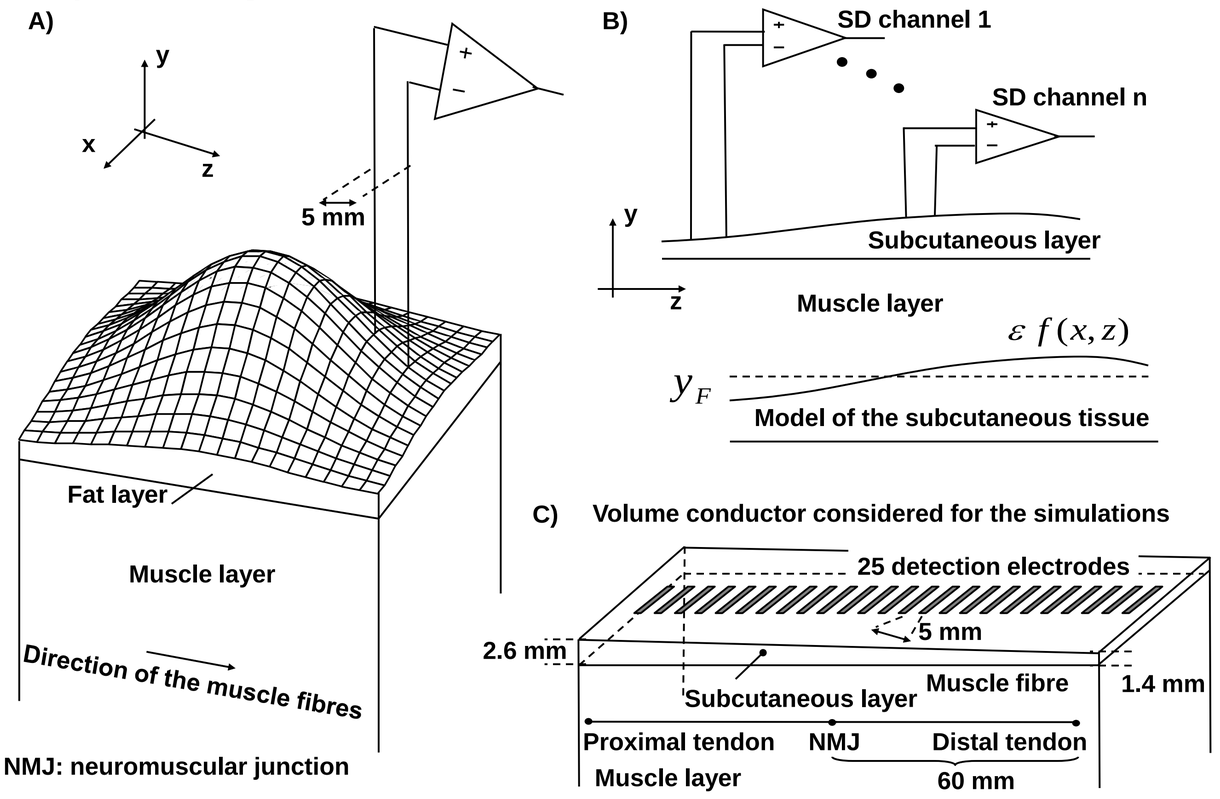


Fig.2 A) Volume conductor with variable thickness of the subcutaneous tissue layer. B) Cross section of the volume conductor. C) Volume conductor with linear variation of subcutaneous tissue layer thickness.

Table 1: Simulated conductivities

Subcutaneous tissue	Muscle: transversal conductivity	Muscle: longitudinal conductivity
$\sigma = 4.3 \cdot 10^{-4}\ S/m$	$\sigma_{MT} = 0.09\ S/m$	$\sigma_{ML} = 0.4\ S/m$

The thickness of the fat layer is divided into two contributions: a constant value and a variable function $G(x,z) = y_F + \varepsilon f(x,z)$

Assumed smooth and of the same magnitude as y_F

Assumed very small $\varepsilon << 1$

A regular perturbation expansion of the potential can be considered

$$\varphi(x,y,z) = \varphi^0(x,y,z) + \varepsilon\varphi^1(x,y,z) + \varepsilon^2\varphi^2(x,y,z)$$

A Taylor series of the boundary condition in terms of ε is also considered

$$\hat{n}(x,z) = \nabla Q\big|_{Q(x,y,x)=0}, \qquad Q(x,y,z) = y - G(x,z) = y - y_F - \varepsilon f(x,z), \qquad \nabla Q = \begin{pmatrix} -\varepsilon f_x \\ 1 \\ -\varepsilon f_z \end{pmatrix}$$

$$0 \equiv \nabla Q \cdot \nabla\varphi\big|_{y=y_F+\varepsilon f(x,z)} = -\varepsilon f_x \varphi_x(x, y_F+\varepsilon f, z) + \varphi_y(x, y_F+\varepsilon f, z) - \varepsilon f_z \varphi_z(x, y_F+\varepsilon f, z) \cong$$
$$= \varphi_y(x,y_F,z) + \varepsilon f \varphi_{yy}(x,y_F,z) - \varepsilon\left(f_x\varphi_x(x,y_F,z) + f_z\varphi_z(x,y_F,z)\right) + O(\varepsilon^2)$$

Equating the same powers in ε, a hierarchic mathematical problem (each problem with solution depending on the solutions of all the preceding ones) is obtained.

Order zero

$$\begin{cases} -\nabla \cdot (\underline{\underline{\sigma}} \nabla \varphi^0) = \delta(x_0, y_0, z_0) \\ \varphi_y^0\big|_{y=y_F} = 0 \end{cases}$$

First order

$$\begin{cases} -\nabla \cdot (\underline{\underline{\sigma}} \nabla \varphi^1) = 0 \\ \varphi_y^1\big|_{y=y_F} = -f\varphi_{yy}^0 + f_x\varphi_x^0 + f_z\varphi_z^0\big|_{y=y_F} = F_1(x,z) \end{cases}$$

Each of these problems is defined in a plane layer volume conductor. The problem of order zero is not homogeneous (a Dirac delta function is considered here to study the impulse response), but has vanishing boundary condition. The other problems are homogeneous, but have a flux term from the boundary. All these problems can be solved analytically transforming the x and z space variables (see Fig. 1) into spatial frequency k_x and k_z by a 2D Fourier transform.

3. RESULTS

A linear variation of the thickness of the subcutaneous tissue layer between 1.4 and 2.6 mm along the direction of the muscle fibres was simulated (Fig. 2C). A linear array with 25 electrodes (5 mm inter-electrode distance), centred over the innervation zone was simulated. Fibres were located in a range of depths 1 – 8 mm and with transversal distances from the detection array in the range -20 mm to 20 mm. Symmetrical muscle fibres with semi-length 60 mm were simulated. Motor unit action potentials (MUAP) were simulated with a spread of neuromuscular junctions and tendons of 8 mm. The number of fibres in the MUs was distributed as an exponential function, with ratio of innervation numbers 20. The distribution of conduction velocity (CV) of the MUs was Gaussian, with mean 4 m/s and standard deviation 0.3 m/s. Interference EMG signals at 80% of maximal voluntary contraction (MVC) were simulated in monopolar and SD configuration (Fig. 3) for 10 random distributions of the MUs within the muscle.

Average rectified value (ARV) and mean frequency (MNF) were estimated from a 5 s portion of simulated signal for each monopolar and SD channel. CV was estimated by a maximum likelihood method from channel pairs. Results are shown in Fig. 3.

Variables can be estimated reliably only far from IZ and tendons. Far from IZ and tendon, ARV and MNF are lower when estimated above a thicker subcutaneous tissue layer (about 10%, 20% variation for ARV and 10%, 5% for MNF estimated from monopolar-SD signals, respectively). CV was not affected by the simulated variation of thickness of the subcutaneous tissue layer.

4. CONCLUSIONS

This work introduces an analytical model of simulation of surface EMG that, together with the other models proposed in the literature, is contributing to understanding the effect of particular conductivity or geometrical properties of the tissues on the recorded signals. Even simulating small variations of subcutaneous thickness, the results provided show that amplitude and spectral variables extracted from EMG are largely affected by the position of the detection point. On the other hand, CV estimated by a maximum likelihood approach from channel pairs is not affected by the thickness of the subcutaneous tissue in the simulated range of variation.

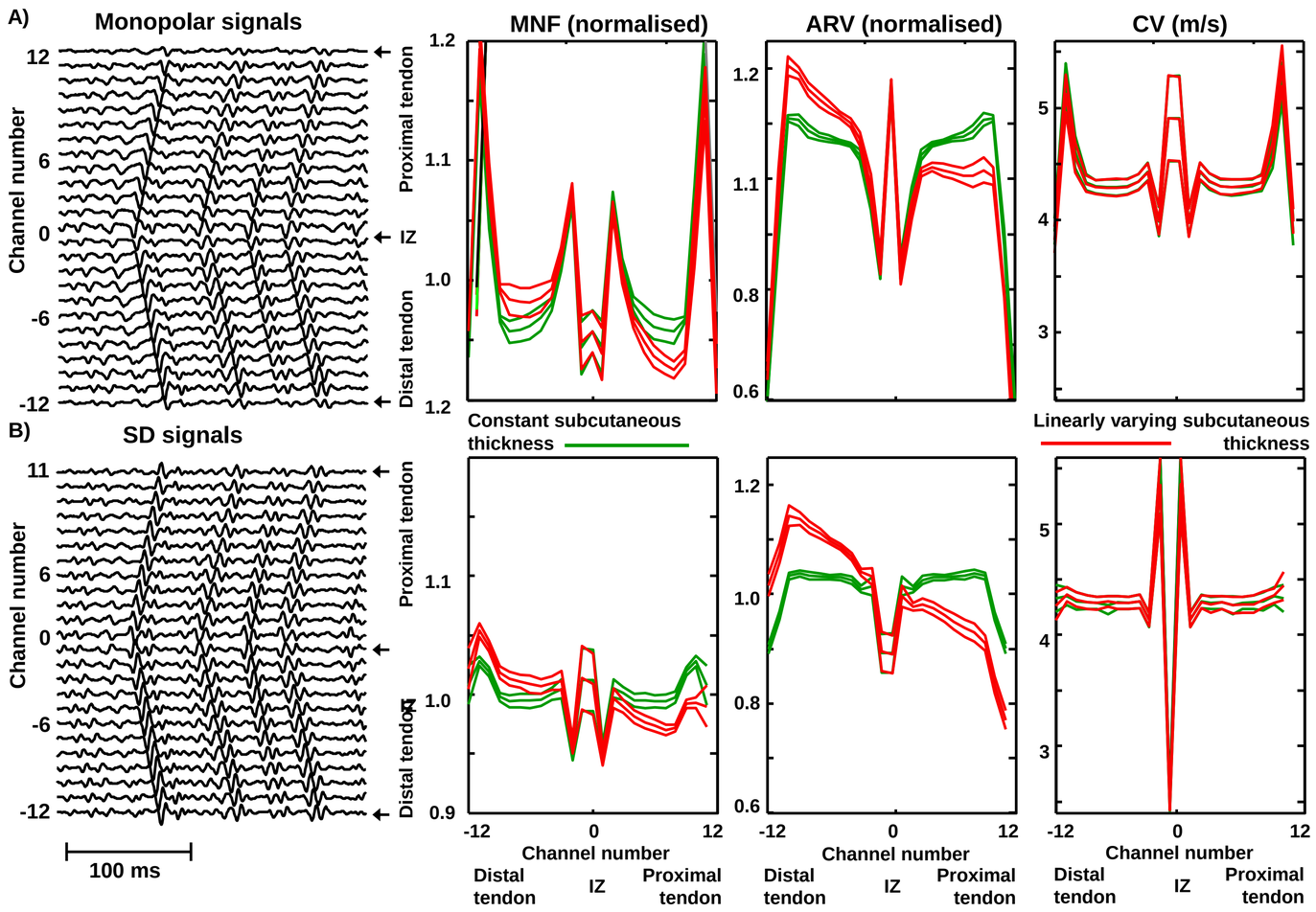


Fig.3 Example of A) monopolar and B) single differential (SD) interference signal and mean ± standard deviation (over 10 simulated distributions of the MUs within the muscle's cross section) of ARV, MNF and CV (between adjacent pairs of channels). ARV and MNF from each simulation were normalised with respect to the mean value across channels.

References

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[2] Farina D., Mesin L., Martina S., Merletti R., A surface EMG generation model with multi-layer cylindrical description of the volume conductor, *IEEE Trans. on Biomed. Eng*. 51: 415-426, 2004.

[3] Mesin L., Farina D., Simulation of surface EMG signals generated by muscle tissues with in-homogeneity due to fiber pinnation, *IEEE Trans. on Biomed. Eng*., 51: 1521-1529, 2004.

[4] Mesin L., Joubert M., Hanekom T., Merletti R., Farina D., A Finite Element Model for Describing the Effect of Muscle Shortening on Surface EMG, *IEEE Trans. on Biomed. Eng.*, 53: 593-600, 2006.

Acknowledgements

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