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

L. Mesin
LISiN, Department of Electronics, Polytechnic of Turin, Italy

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 in the literature [1] to simulated surface EMG. In spite of the higher flexibility of numerical methods, analytical solutions are valuable to check the accuracy of numerical methods, to reduce the computational time and to determine the theoretical dependence of the solution on specific parameters of the system.

Surface EMG can be described as a propagating wave in space invariant volume conductors. This allows reducing the simulation time and considering the detection channels as spatial filters. Recent works investigated also non space invariant volume conductors, indicating the effect of tissue inhomogeneity or complex geometry on the simulated EMG signal. Also small perturbations from space invariant volume conductors can have important effects on surface EMG and on EMG variables. Models can provide indication on the effect of a specific perturbation from the ideal space invariant volume conductor. This work introduces a new analytical model for the generation of surface EMG signals from a non space invariant volume conductor constituted by a planar muscle and a subcutaneous tissue with variable thickness.

METHODS

Tissues are modelled as volume conductors in stationary conditions. The electric potential is obtained by solving an electrostatics problem with isolation condition at the surface $-\nabla \cdot (\underline{\underline{\sigma}} \nabla \varphi) = I$, where φ is the electric potential, I is the source current density, and $\underline{\underline{\sigma}}$ the conductivity tensor. A two layer volume conductor describing subcutaneous tissue and muscle tissues is considered. The subcutaneous tissue is assumed isotropic with conductivity $\sigma = 4.3 \cdot 10^{-4}$ S/m. Muscle is anisotropic, with transversal conductivity $\sigma_{MT} = 0.09$ S/m and longitudinal conductivity $\sigma_{ML} = 0.4$ S/m. Muscle layer is a hemi-space. Subcutaneous tissue layer is placed over the muscle and has a variable thickness, described by a function $y = G(x,z)$ (Figure 1). The thickness of the subcutaneous tissue is divided into a constant value y_F and a function $f(x,z)$ (smooth and of the same magnitude as y_F) multiplied by parameter $\varepsilon < 1$, so that small variations with respect to the constant thickness y_F are considered. An approximate impulse response is obtained analytically by expanding the function $G(x,z)$ describing the subcutaneous tissue surface

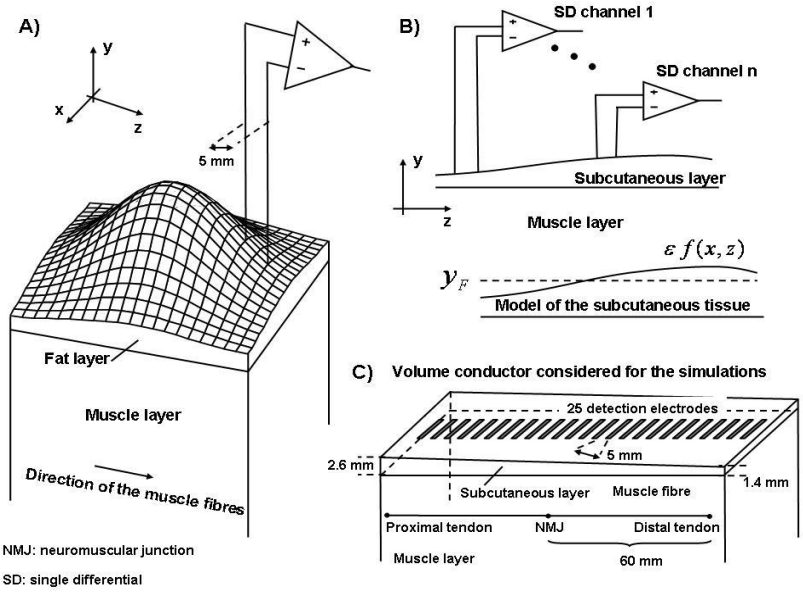

Fig.1 A) Volume conductor with variable thickness of the subcutaneous tissue layer. B) Cross section of the volume conductor. C) Volume conductor with linear variation of subcutaneous tissue layer thickness

and the solution as a series of powers of ε. A hierarchy of Poisson problems is obtained for a planar geometry, for which a solution can be found. Series truncation to a second order approximation was considered.

RESULTS AND DISCUSSION

As an example of application, 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 considered. A linear array with 25 electrodes (5 mm inter-electrode distance), centred over the innervation zone was simulated (Figure 1C). 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 (Figure 2). 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 [2], 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 were simulated for 10 random distributions of the MUs within the muscle (80% MVC) using the model in [2].

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 (Figure 2). 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 estimated from monopolar-SD signals, respectively; 10%, 5% variation for MNF estimated from monopolar-SD signals, respectively). CV was not affected by the simulated variation of thickness of the subcutaneous tissue layer.

In conclusion, 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.

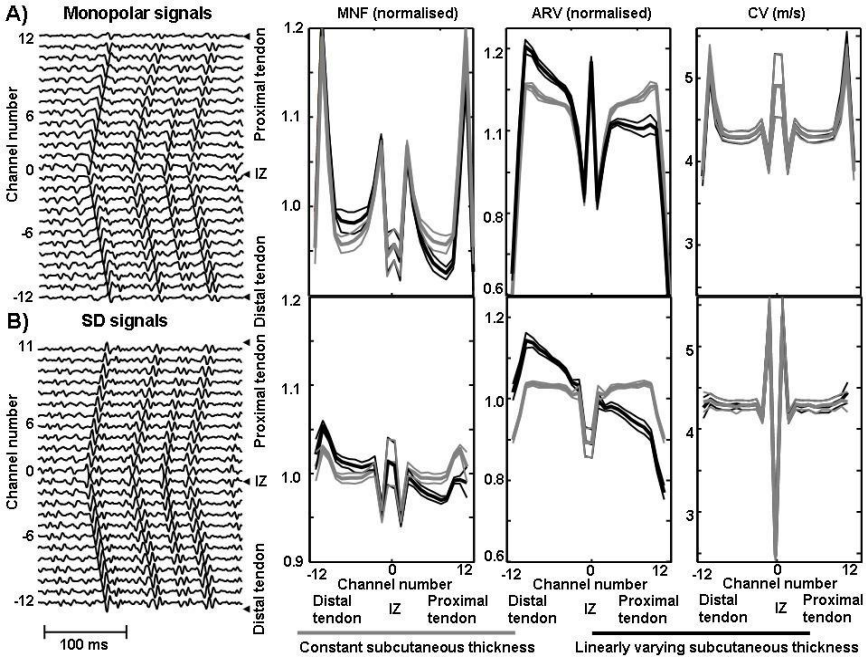

Fig.2 Example of A) monopolar and B) single differential (SD) interference signal and mean ± standard deviation (over 10 simulated distributions of the MUs within the muscle's cross section) of ARV, MNF and CV. ARV and MNF from each simulation were normalised with respect to the mean value across channels.

REFERENCES

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