Mathematically modelling surface EMG signals

Mylena Mordhorst1,2,*, Thomas Heidlauf1,2, and Oliver Röhrle1,2

1 Institute of Applied Mechanics (CE), University of Stuttgart, Pfaffenwaldring 7, 70569 Stuttgart, Germany
2 Stuttgart Research Centre for Simulation Technology (SimTech), University of Stuttgart, 70569 Stuttgart, Germany

Comprehension and correct interpretation of EMG signals and their generation could still be well improved. Computational models that can predict the EMG signal resulting from realistic motor unit recruitment as well as the underlying biophysical processes of single skeletal muscle fibres are therefore highly desirable. Having such a model available, one can test, verify and improve algorithms determining motor unit recruitment. Here, we present a three-dimensional, continuum-based, forward model that is able to produce a virtual EMG signal based on the underlying biophysical principles of skeletal muscle fibre activation. The result is a virtual EMG signal for complex and realistic geometries that may even undergo deformations as in the case of dynamic contractions.

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1 Motivation and introduction

Due to several limitations of experimental EMG recordings, computational models that allow to investigate the relationship between the measured EMG signal and the underlying biophysical processes were developed in the last decades. Many of the existing models are analytical models, see for example [1, 2], that are restricted to simple geometries and static solutions. This drawback does not apply to numerical models applying the finite element method (e.g. [3]). The model introduced in the following, is based on the idea of coupling the cellular electrophysiology with the macroscopic equations of finite elasticity, which was first presented in [4].

2 Methods

The source of the EMG signal is the electrical activity of a single skeletal muscle fibre. The muscle fibres belonging to one motor unit (MU) are stimulated at the neuromuscular junction through their corresponding motoneuron. This initiates an action potential (AP) propagating along each fibre from the neuromuscular junction towards the ends. Our virtual EMG model basically consists of two parts. The first part is the calculation of the AP, i.e. the change of the transmembrane potential in time, along a single skeletal muscle fibre. For this task, we make use of the chemoelectromechanical skeletal muscle model from [5], which uses a detailed biophysical model and the monodomain equation to solve for the AP propagation along a single fibre. In a second step, the distribution and the propagation of the electric potential inside the muscle and in the fat/skin layer has to be computed, in order to obtain the (surface) EMG signal.

A summary of the governing equations for step two can be seen in Fig. 1. The equation that has to be solved for the potential layer has to be computed, in order to obtain the (surface) EMG signal.

![Fig. 1: Summary of the governing equations for the computation of a virtual EMG signal: The transmembrane potential, \( V_m \), is precomputed in the whole muscle domain for several timesteps, \( t_k \). The result is introduced into the right-hand side of the extracellular bidomain equation. \( \sigma_i, \sigma_e \), and \( \sigma_o \) are the intra- and extracellular muscle fibre conductivities and the conductivity of the fat/skin layer. The extracellular potential is denoted by \( \phi_e \), while \( \phi_o \) is the electric potential in the fat/skin layer. Further, \( \Omega := \Omega^M \cup \Omega^B \) and \( \phi \) is either \( \phi_e \) or \( \phi_o \), depending on the region \( \Omega \).](image-url)
These equations are accompanied by boundary conditions, which assure a continuous potential across the muscle-body-boundary (c.f. Eq. (3), (4)) and make sure that no potential flows out of the body region (c.f. Eq. (5)). Furthermore, to account for a reference potential of 0 mV, a zero-mean condition is imposed on the calculated surface potential (c.f. Eq. (6)). For more details on these steps, see [6, 7]. The linear system of equations (not including the boundary conditions) is given as

$$\begin{bmatrix} K_{i+e} & 0 \\ 0 & K_o \end{bmatrix} \begin{bmatrix} \phi_e \\ \phi_o \end{bmatrix} + \begin{bmatrix} K, V_m \end{bmatrix} = \begin{bmatrix} 0 \\ 0 \end{bmatrix},$$

where $K_{i+e}, K_i, K_o$ are global matrices assembled from element matrices that depend on the conductivities $\sigma_{i+e}(x), \sigma_i(x), \sigma_o(x)$ respectively. Here, $\phi_e, \phi_o, V_m$ are the vectors of the nodal values for the potentials $\phi_e, \phi_o, V_m$.

### 3 Results and Discussion

The discretised equations were implemented into the open source code OpenCMISS (c.f. [8]). To test the method, a tissue block consisting of an anisotropic muscle layer with 100 embedded fibres covered by an isotropic fat/skin layer is created. The tissue conductivities (in $\text{mS cm}^{-1}$) are assumed to be constant and are given by

- $\sigma_i := \text{diag}(8.93, 0.893, 0.893)$,
- $\sigma_e := \text{diag}(6.7, 6.7, 6.7)$,
- $\sigma_o := \text{diag}(0.4, 0.4, 0.4)$.

The 2D EMG signal on the skin surface for one time instant resulting from a random fibre stimulation can be seen in Fig. 2. Plotting the calculated potential at one node over time yields the 1D interferential EMG signal at that position (c.f. Fig. 3).

![Fig. 2: The 2D virtual surface EMG signal. The potential distribution on the surface is represented by colours. The black circles (○) represent possible electrode locations.](image)

![Fig. 3: The 1D interferential EMG signal (raw and rectified) taken at the muscle surface at position • marked in Fig. 2.](image)

Furthermore, the simulation can be performed on a deforming domain, simulating the electrical activity during an activation induced dynamic contraction. As a next step, the model should be verified with a realistic testing case, e.g. biceps brachii under various isometric contractions. From then on, the model could be used for several case studies on EMG signals, e.g. studies on signal depth and the influence of the fat skin thickness and conductivity on the obtained potential.

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### References