A 3D transient model for transcutaneous functional electrical stimulation

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Abstract

Knowing the 3D current and potential distribution is important for more precise stimulation in transcutaneous functional electrical stimulation. Currently static models can describe the effect of an amplitude change of the stimulation, but the result is the same for different pulse durations. We have developed a transient quasistatic finite element (FE) model that simulates the potential distribution inside the skin, fat, muscle and bone layers of the upper arm. The different tissues are defined by their conductive and dielectric properties. The inclusion of the dielectric properties accounts for capacitive effects in the different tissue layers. This factor has been neglected in earlier studies where only static simulations using the Poisson equation were performed. Our simulation results compared well to external and intramuscular voltage measurements that were performed on three human volunteers. The experimental results showed that the dielectric properties of the skin have to be included in simulations whereas the dielectric properties of the muscle might be neglected.

1. INTRODUCTION

In electrical stimulation the pulse duration has an influence on the muscle activation. This is most probably due to capacitive effects either in the nerve membrane or in the volume conductor. The capacitive effects of the nerve membrane are included in the widely used Huxley type and transmission line models. However, all existing models that the authors are aware of consider a static field distribution inside the volume conductor, although the material properties of skin, fat and muscle would suggest a transient solution. With this research we want to find out if the capacitive effects of the volume conductor (human tissues) affect the muscle activation during functional electrical stimulation.

Simulating static fields [1, 2] means that when a signal is applied the potential in the medium can be assumed to be in steady state instantaneously. Most authors justify the static approach with [3] who gave a condition when capacitive effects can be neglected, which is

$$\frac{\omega \varepsilon_0 \varepsilon_r}{\sigma} << 1.$$

Measurements of conductivities and permittivities of different tissue types show that there exist values for which this formula does not hold [4]. As a result the capacitive effects, which are a consequence of the dielectric properties (permittivity) cannot be neglected in skin, fat and muscle tissue. It has been shown in other fields (e.g. simulation of EMG signals) that capacitive effects can have an influence [5]. Therefore, we developed a finite element model of the human upper arm for transcutaneous electrical stimulation that also incorporates the capacitive effects of the most important tissues. With this model we wanted to find out in which cases the dielectric effects of the volume conductor have an influence on the activation of the nerves and muscles.

2. METHODS

2.1. Transient Finite Element Model

The geometry of the human upper arm was modelled by concentric cylinders for the skin (1.5 mm), fat (8.5 mm), muscle (27.5 mm), cortical bone (6 mm) and bone marrow (6.5 mm) layers. Each layer was described using the conductivity σ and also the permittivity ε. A large range of values for σ and ε at each layer have been presented in literature.

FE Model	Muscle (longitudinal)		Muscle (transverse)		Skin	
	ε_r	σ	ε_r	σ	ε_r	σ
A	1.2e5	0.315	4e4	0.105	1e4	2e-4
B	1.2e5	0.55	4e4	0.183	1e4	1.23e-4
C	1.2e5	0.16	4e4	0.053	8e3	1.66e-4
D	1.2e5	1	4e4	0.333	6e3	2.5e-4
E	1.31e6	0.315	5.2e5	0.105	6e3	3.33e-4
F	1.31e6	0.08	5.2e5	0.027	6e3	3.33e-4

Table 1: This table shows the tissue parameters that were used for the FE simulation.

Therefore, the FE simulation was performed for different combinations of the values in order to

establish the models sensitivities to the different properties.

The used properties of the different layers are shown in Table 1 and are in agreement with [4]. Fat ($\sigma = 0.03$, $\varepsilon_r = 25000$), bone ($\sigma = 0.02$, $\varepsilon_r = 3000$) and bone marrow ($\sigma = 0.08$, $\varepsilon_r = 10000$) were held constant because we were mainly interested in the influence due to the skin and muscle properties.

Wave propagation and inductive effects were assumed to be negligible [3].

The time-dependent differential equation for the scalar potential was described by

$$-\nabla\cdot\left([\sigma]\nabla V\right)-\nabla\cdot\left([\varepsilon]\nabla\frac{\partial V}{\partial t}\right)=0\,.$$

The mesh of 3D elements and the calculations of the potential distribution were produced with the commercial package EMAG from ANSYS (see Figure 1). In FE modelling it is very important that the mesh elements are small enough to ensure an accurate solution. Therefore, the mesh-size was refined until no significant change could be found in the transient solution at every time point of the simulation.

We implemented different shapes and sizes of electrodes into the finite element model, which were assigned the properties of gold. For comparison with our measurements square 5 cm electrodes were chosen and the distance between the two electrodes was 6 cm. Monopolar constant current pulses with a pulse duration of 100 µs and 1 ms and an amplitude of 9 mA were applied to the electrodes of the model.

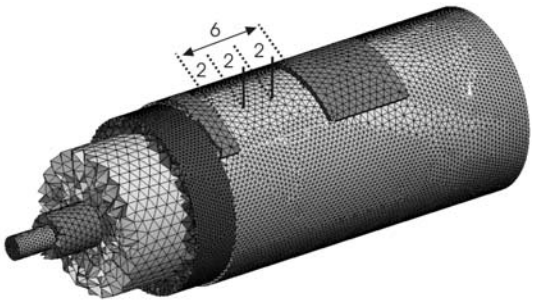

Figure 1: Finite Element Mesh with different tissue layers and the exciting electrodes. Recording needle electrodes between the transcutaneous electrodes are shown for clarity (not included in FE model).

2.2. Measurements

We compared our transient finite element model with experimental data that was obtained in three healthy male subjects. The size and distance of the electrodes were the same as in the FE model (squares of 5 cm, distance 6 cm). A standard current regulated stimulator was used for the electrical stimulation [6]. As with the FE model monopolar current pulses with a pulse duration of 100 µs and 1 ms and amplitude of 9 mA were used for the stimulation.

We were interested in the potentials inside the muscle and at the electrodes. The transient potentials inside the muscle were measured with needle electrodes from Medtronic (9013R0252). Two needle electrodes were inserted 2 cm into the biceps muscle with a spacing of 2 cm (see Figure 1). In addition we also measured the current and external voltage between the two stimulating electrodes. This external measurement showed the transient response of the whole system (electrode, skin, fat, muscle) to a monopolar current pulse of the stimulator.

3. RESULTS

The two charts in Figure 2 show the result of a simulation using the transient finite element model for the material parameters from Table 1.

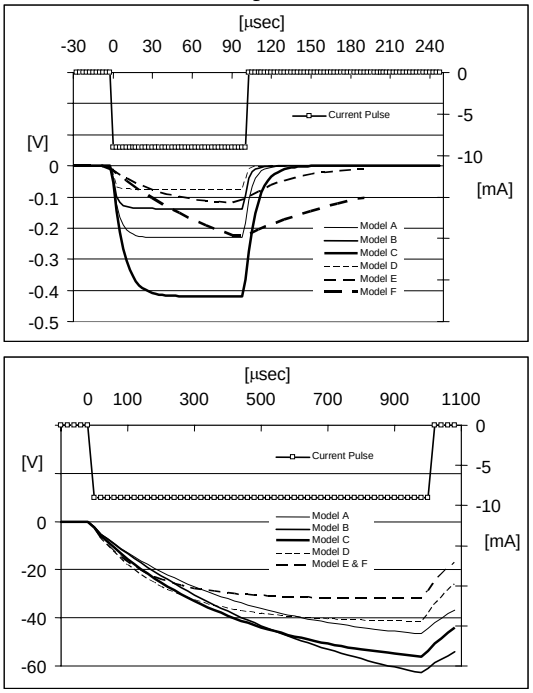

Figure 2: The two graphs show the simulations for the tissue parameters from Table 1. The upper chart contains the intramuscular voltage and the lower chart the voltage between the stimulation electrodes. Both charts additionally show the exciting constant current pulse.

The top chart depicts the intramuscular voltages at equidistant time steps of 2.5 µs. The simulated potential was interpolated at the same two points inside the muscle as recorded in the validation experiments 2 cm under the skin aligned in axial direction with the cylinder (see Figure 1). The simulation results showed that

there exist published values for the tissue parameters that have a slow rise time of the potential inside the muscle (Model E and F). This slow rise time is due to the high permittivity and it would result in a higher threshold for the nerve activation. A static model would overestimate the nerve activation.
The lower chart in Figure 2 shows the voltage between the electrodes during a monopolar current pulse. It shows the high capacitance that is always present in the skin layer.
The two charts in Figure 3 show the internal and the external voltage measurements during a monopolar constant current pulse that were performed in three human volunteers. Model D (see Figure 2) correlates best with the experiments.

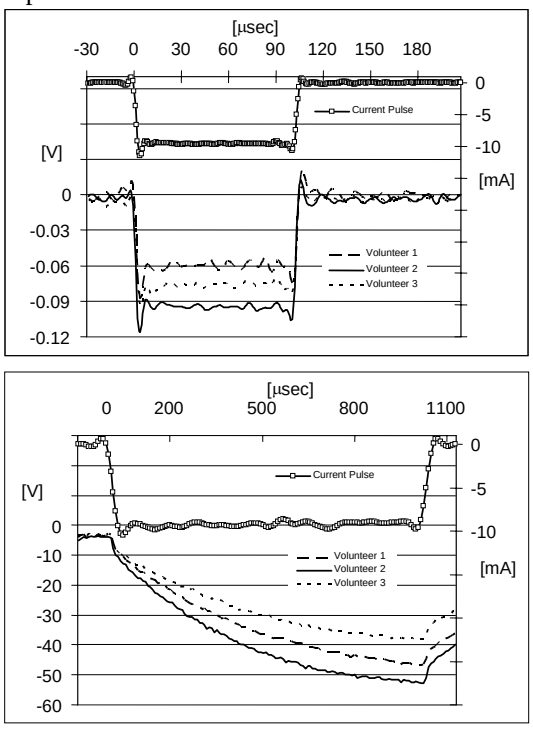

Figure 3: The two figures show measurements that were performed in three human volunteers. The top chart contains the intramuscular voltage measurements using needle electrodes and the lower chart the voltage over the stimulation electrodes during a pulse. Both charts additionally show the exciting constant current pulse.

4. DISCUSSION AND CONCLUSIONS

A transient quasistatic finite element (FE) model was developed to simulate the evolution of the electrical scalar potential inside the human body over time. Previous modelling approaches have neglected the transients [3]. Our model was validated with internal and external voltage measurements in the upper arm of three human volunteers.
The experimental measurements showed that the capacitive currents in the skin were bigger than the conductive currents when a short current pulse (less than 500 µs duration) was applied. This can also be observed with all parameter combinations from Table 1 in our finite element model. Although we are always using constant current stimulation we want to point out that when simulating constant voltage stimulation it is necessary to use a transient simulation of the potentials instead of a static model. Applying a constant voltage to the electrodes results in a falling voltage at the muscle because the longer the pulse lasts the more voltage gets lost over the skin.
Inside the muscle there exist combinations of published tissue parameters that result in short rise times of the potential but there also exist combinations producing long rise times in the FE model. Our intramuscular needle experiments showed consistently very short rise times inside the muscle (see Figure 3). This suggests that the capacitive effects could be neglected inside the muscle. However, a big range of the published values are like Model E and F, which lead to slow rise times in the muscle (Figure 2). Only the rise time of Model D resembles our measured values.
Yet it is not clear where these discrepancies between most published tissue parameters used for simulations and our experiments come from.

References

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