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Numerical Simulation of the Energy Distribution in Biological Tissues During Electrical Stimulation

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Abstract: Functional Electrical Stimulation (FES) is used for activation of nerves and muscles in order to restore their normal functions after an injury or stroke. Taking into consideration that electrical stimulation does not directly activate neural tissues, our intention was to show how this power is distributed in the tissues. In order to estimate the power (energy) distribution in the natural multilayer system of different tissues (skin, fat, muscle, and bone) we considered their respective geometries and electrical properties. Simulation was performed using a tissue model based on a finite element method. We present the energy distribution for each layer, and show that the largest portion of the energy is lost in the top skin layer, and that only a fraction reaches the targeted neural tissue.

Keywords: Functional Electrical Stimulation (FES), Finite Element Method Modelling, Energy distribution, Tissue, Surface multi-pad electrodes.

1 Introduction

One of the most promising approaches for restoration of motor function in stroke, and in complete and incomplete, paraplegic and tetraplegic patients with permanent limb impairment, is Functional Electrical Stimulation (FES). Although the first FES applications were described almost 50 years ago, with considerable technical improvements made since, the FES technology has had very limited impact on the rehabilitation of paraplegic and tetraplegic patients so far. The present FES treatments, combined with conventional occupational and physical therapy, still remain the most effective approach in rehabilitation for restoration of grasping and walking [1 – 3].

Action potentials related to FES are initiated via electrodes, mostly applied to the surface of the skin. The different types of electrodes used for electrical stimulation, together with their properties, are reviewed in [4]. Recent investigation [5] has shown that the electric properties of the electrode-tissue

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interface have a great influence on current density distribution in the tissues. Various methods of analysis, and modification of the electrode-tissue interface, have been proposed in order to improve the distribution of current densities underneath the surface electrodes, but prediction of the electrical behaviour of biological tissues such as skin, fat, muscle and bone is difficult because of their heterogeneity and anisotropy [6]. As knowledge of the current distribution within these excitable tissues is a significant factor for prediction and control of the muscle output, it is important to further investigate electrical effects that appear locally, thus highlighting a potential region of interest for FES.

In [7], a hybrid scheme was used for the calculation of the intramuscular three-dimensional current density distribution and potential field generated by transcutaneous electrical stimulation. It was shown that current density was not dependent on the electrode size; however, the inclusion of the bone–fascia layer significantly increased the intramuscular current density slope. This shows that in modelling of biological tissues, anatomical characteristics must be taken into account, and realistic parameters need to be employed.

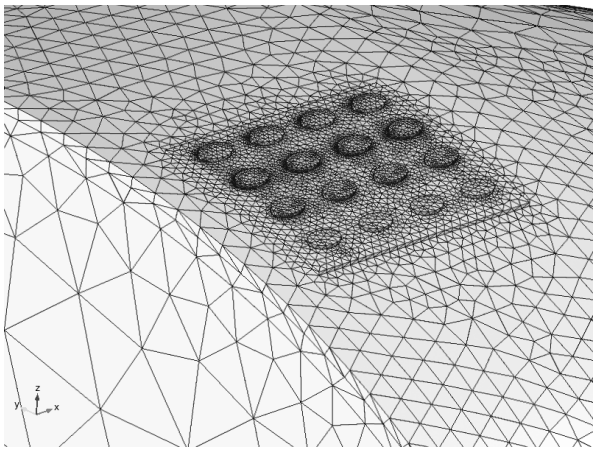
This paper presents the results of numerical simulations of electrical stimulation in layered biological tissue via transcutaneous electrodes. In order to be able to experimentally validate our model in the future, we created a model from porcine tissue instead of human tissue. It is known that all of the major structures found in humans are present in pigs. Furthermore, pigs have the same muscles as humans, with some small variations in the size and location of a subset of these related to the fact that pigs are quadrupedal and humans are bipedal. In the hind limb, a pig has the same muscles as a human in the major thigh muscle groups: quadriceps femoris and the hamstrings [8]. For this reason, we developed our simulations on a model based on a porcine hind leg. The aim of the study was to estimate the energy distribution in the natural multilayer system of different tissues (skin, fat, muscle, and bone), taking into account their respective thicknesses and impedance.

2 Methods

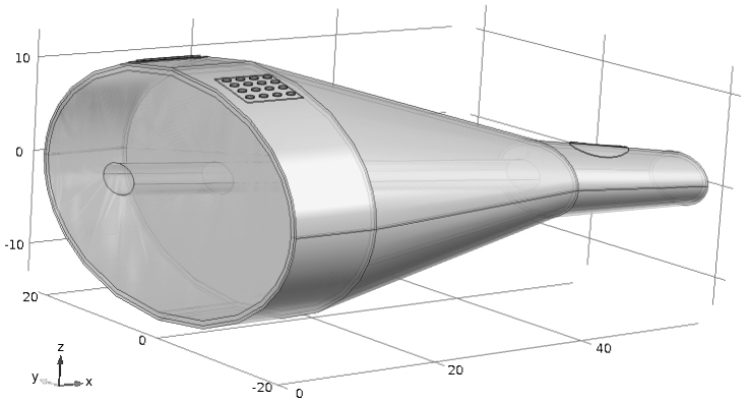
The numerical model was used to simulate the energy distribution in porcine legs during electrical stimulation. As we were observing the loss of power in the skin, fat, muscle, and bone due to the Joule effect, a stationary regime was employed. Furthermore, it was shown that all the tissue capacitances can be safely disregarded except for the capacitance of skin, and consequently that all the tissues below the skin reach a stationary regime, and maintain it during the stimulation pulse (300 μs).

For modelling we used COMSOL Multiphysics 4.2 software suite, which employs a finite element method. This software is able to generate a mesh based on a defined geometry (Fig. 1a), set boundary conditions, iteratively solve a system of equations and visualise the results. In order to calculate electrical potentials in our model we employed the continuity equation:

$$-\nabla\cdot\left(\left[\sigma\right]\nabla V\right)-\nabla\cdot\left(\left[\varepsilon\right]\nabla\frac{\partial V}{\partial t}\right)=0\,. \tag{1}$$



(a)



(b)

Fig. 1 – a) *Mesh grid of the numerical model. Grid is fine in areas where the gradient of current density is expected to be greater and* b) 3D *model of a porcine leg with electrode placement.*

The model of a porcine leg (Fig. 1b) was based on pictures and cross-sections found on the University of Nebraska – Lincoln website [9], while the

dimensions were taken from [10]. The model included the following tissue types: skin, fat, muscle, and bone. We made four models to compare different combinations of skin and muscle tissue parameters (models A1, A2, B1, and B2). Different muscle tissue parameters were used to check the variability in energy distribution due to the direction of the muscle fibres (along or across the porcine leg). Dielectric properties of all the tissues, except for the muscle tissue of Model 2, were calculated according to the data and equations found in Gabriel et al. [11, 12], see **Table 1**. Dielectric properties for the muscle tissue in Model 2 were taken from [13].

Table 1

Dielectric properties of the tissues.

Tissue name	Conductivity [S/m]
Wet Skin	Model A: 0.00333 Model B: 0.00042719
Fat	0.019555
Muscle	Model 1: 0.23329 Model 2: 0.09
Cortical Bone	0.020055

Three electrodes were modelled to represent the main stimulation equipment used: two multi-pad electrodes with 16 individual pads (described in [14]) that acted as cathodes, and one single elliptical anode (5.5×7) cm. Multi-pad electrodes were placed on the upper part of the model of the porcine leg (effectively, on the thigh), while the anode was positioned on the lower part of the model (in the vicinity of the knee), see Fig. 1b. Intensity of DC stimulation current was set to $I = 10$ mA. Modelled electrodes, their placement, and values of the current employed are those typically used in clinical FES applications.

For the parameters in **Table 1**, the simulation produced electrical potential isosurfaces on which we based the energy distribution in the model. Active volume, in which we estimated power loss, was between the first (V1) and the last (V2) isosurface, with the electrical potential that is inside the tissue of interest. Estimated power loss was then calculated using equation (2):

$$P = (V_2 - V_1)I \,. \tag{2}$$

3 Results

For all models described above, we ran simulations of electrical stimulation with a current intensity $I = 10$ mA. The results are shown in Figs. 2, 3 and 4.

Calculated current density streamlines for Model A1 are shown in Fig. 2.

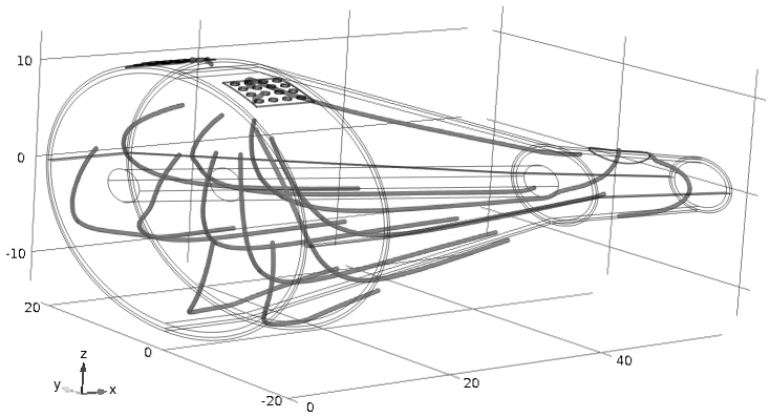


Fig. 2 – *Current density streamlines for Model* A1.

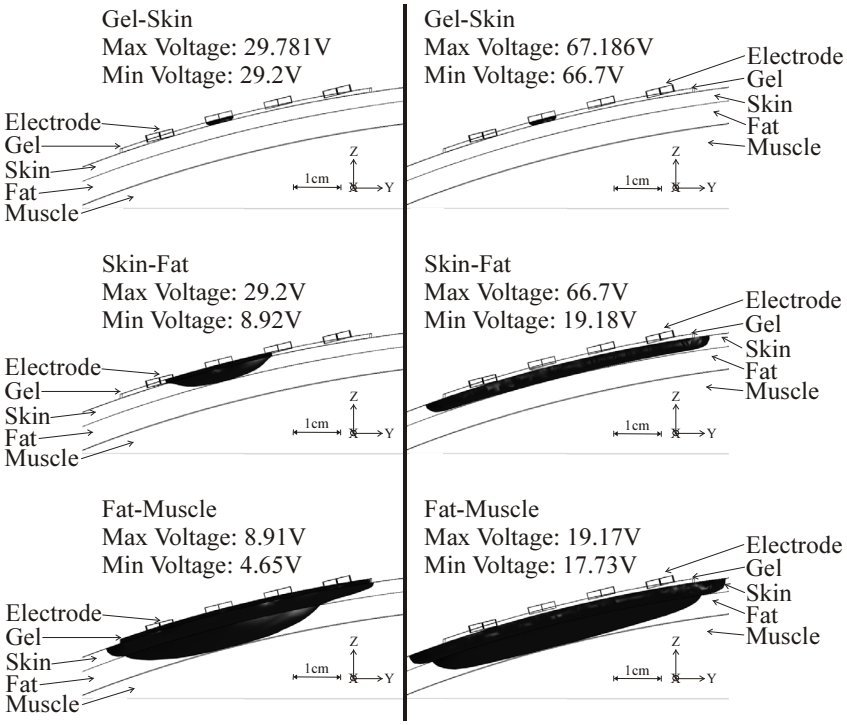


Fig. 3 – *Voltage boundary states of different tissues;*
Model A1 *– left column, Model* B1 *– right column,*
First row – voltage drop inside skin tissue,
Second row – voltage drop inside fat tissue,
Third row – voltage drop inside muscle tissue.

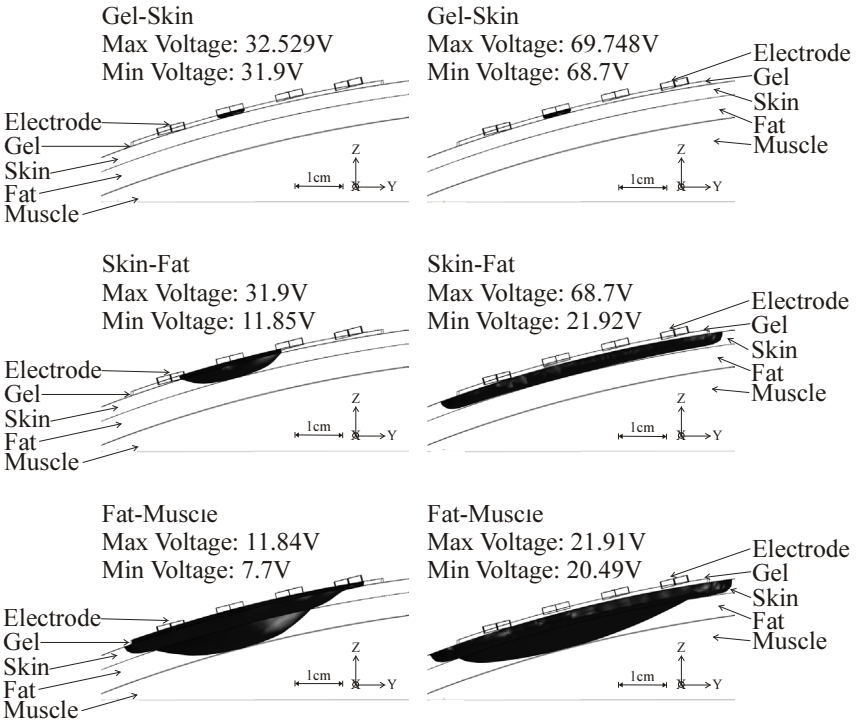


Fig. 4 – *Voltage boundary states in different tissues; Model* A2 *– left column, Model* B2 *– right column, First row – voltage drop inside skin tissue, Second row – voltage drop inside fat tissue, Third row – voltage drop inside muscle tissue.*

The most significant isosurfaces that we used to calculate power losses in tissues are shown in Figs. 3 and 4. Maximum voltage in Model A1 was 30 V and in Model B1 it was 67 V. Considering that the intensity of current stimulation was the same for both models (I = 10 mA), total power spent for stimulation in Model B1 is more than twice the power spent in Model A1.

Results from the numerical models of energy distribution in biological tissues during electrical stimulation are presented in **Table 2**.

Table 3 presents sums of power losses at the gel-skin interface beneath the anode and cathode. The addends were calculated using equation (2), where V_2 was the maximum voltage potential for the given model and V_1 was the voltage potential at the gel-skin interface beneath the multi-pad electrode if the power loss is calculated beneath the cathode, else V_2 was the voltage potential at the gel-skin interface beneath the anode and $V_1 = 0$.

Presented results from **Tables 2** and **3** are based on voltage values extracted from the COMSOL graphical interface. Estimated error for voltage readings is around 0.2V which is ±0.7% for Model A1, ±0.6% for Model A2 and ±0.3% for Models B1 and B2.

Table 2
Energy distribution in biological tissues during electrical stimulation.

Tissue	Power [%]	Power per Volume [W/cm3]	Power [%]	Power per Volume [W/cm3]
	Model A1: Skin = 0.00333 [S/m] Muscle = 0.23329 [S/m]		Model B1: Skin = 0.00042719 [S/m] Muscle = 0.23329 [S/m]	
Skin (Cathode)	67.43	0.2933	70.13	0.0370
Fat (Cathode)	14.30	0.0109	2.14	0.0007
Muscle	7.49	2.5×10^{-6}	2.88	2.2×10^{-6}
Skin (Anode)	6.01	0.0018	22.51	0.0208
Fat (Anode)	1.68	0.0003	0.80	0.0004
	Model A2: Skin = 0.00333 [S/m] Muscle = 0.09 [S/m]		Model B2: Skin = 0.00042719 [S/m] Muscle = 0.09 [S/m]	
Skin (Cathode)	61.17	0.3057	67.07	0.0369
Fat (Cathode)	12.29	0.0160	2.04	0.0023
Muscle	16.66	6.2×10^{-6}	6.99	5.6×10^{-6}
Skin (Anode)	5.11	0.0017	21.43	0.0138
Fat (Anode)	1.51	0.0004	0.76	0.0004

Table 3
Power loss at gel-skin interface.

Model	Power [% of the total power]
A1	2.28
A2	2.85
B1	0.87
B2	1.65

4 Discussion and Conclusion

Functional electric stimulation can help paraplegic, tetraplegic and post-stroke patients regain some of their motor functions. In order to use FES effectively, we must understand the way that current and energy propagate through biological tissues during electrical stimulation. We built a model of a porcine hind leg and ran simulations of FES, observing energy distributions in the skin, fat, muscle, and bone of the model.

Presented simulation model proved that the majority of power is lost in the skin tissue beneath the cathode (~70% for Model A1). Depending on the literature source of tissue parameters, there are global (total energy dissipation) and local (energy distribution in tissues) changes in numerical solution. The major difference between Model A and Model B is that current density is greater in the skin tissue of Model A, thus a significantly smaller volume of skin is affected. This is due to the better conductivity of skin tissue in Model A, which is closer to values that could be found in various publications and were observed in work with patients. With a lower conductivity, a matrix electrode loses its selectivity and thus all its pads behave like one big electrode.

Comparing the Models A2 and B2, with the Models A1 and B1, data in **Table 2** shows that muscle tissue parameters significantly affect the energy deposition in muscle tissue as a local parameter, but have a little overall influence on power loss in skin and fat tissue (~5% max). This proves that the direction of muscle fibres is important for generating accurate models of porcine hind legs and that anisotropic dielectric properties of muscle tissue should be used instead of isotropic ones.

Future work would be to identify parameters of the model based on *in vitro* and *in vivo* experiments with pigs, and by comparing these results with the literature. Our main goal is to build a model that could be used for various stimulation simulations, with different electrodes and with different parameters of stimulation.

5 Acknowledgement

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