2D FINITE ELEMENT MODELLING OF CURRENT DENSITY DISTRIBUTION DURING THE APPLICATION OF ELECTRICAL STIMULATION USING SURFACE ARRAYS

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Abstract: This paper presents the development of two 2D models used to simulate the current density distribution (CDD) under the tissue stimulated by two different electrode array configurations. Finite Element Analysis (FEA) was applied to obtain the CDD in quasi-static conditions. For both models, two different conductive gels were assessed. The results suggested that improved selectivity and more uniform CDD can be achieved when using the less conductive gel, combined with an electrode array of 2mm wide elements separated by 4 mm. Hence, using such a configuration not only would selectivity be improved, but pain would be minimised. This work emphasised the influence of parameters such as electrode size, inter-electrode spacing, electrolyte conductivity and thickness, in the effectiveness of the application of electrical stimulation using surface electrode arrays. Although the models presented here proposed values for such parameters, further experiments in vivo are being carried out to corroborate the results obtained from the models.

Introduction

Functional Electrical Stimulation using surface electrodes was proposed in 1961 as an orthotic alternative for drop foot correction in hemiplegic subjects [1]. Stimuli are typically delivered by means of a pulse generator connected to one pair of electrodes placed over the skin of the affected limb as follows: one situated over the common peroneal nerve as it passes below the head of the fibula and the other close to the motor point of the tibialis anterior [2]. This allows the generation of hip and knee flexion, as well as ankle dorsiflexion with balanced eversion, to allow the foot to clear the ground when electrical stimulation is activated during the swing phase of the affected leg.

In order to obtain an acceptable response care must be taken in the location of the electrodes. This is a time-consuming task which demands an additional effort by the patient on a daily basis and has been reported as the main cause of disuse of these systems [3]. Surface electrode arrays have been proposed as an alternative to solve this problem [4-6]. However, aspects regarding the necessary number of electrodes, inter-electrode separation, electrode size and shape, and appropriate parameters for the electrode-skin interface still need to be addressed.

Different mathematical models have been proposed as tools for designing surface electrodes. Panescu et al. [7] stated that the design of optimal electrodes for surface electrical stimulation demands the investigation of the current flow across the skin. They developed a 2D Finite Element Model of the electrode-electrolyte-skin interface for one active electrode. This model considered the time variant properties and non-uniform characteristic of the skin while applying different amplitudes of voltage. Panescu and colleagues concluded that high resistivity electrolytes lead to reduction of pain. They highlighted the importance of using this kind of modelling for designing electrodes intended for transcutaneous electrical stimulation.

Later, Livshitz et al. [8] emphasised the importance of establishing the CDD in excitable tissue for predicting the muscle output when electrical stimulation is applied using more than one electrode. A hybrid approach using an image series and moment method was implemented for the calculation of the three dimensional intramuscular CDD and potential field provoked by any electrode forming part of an electrode array. This model allowed the study of the influence of the electrical characteristics of tissue, electrode positioning, electrode size and separation on the CDD. It was reported that uniform current density distribution (CDD) will lead to a reduction in pain and improved selectivity [7, 8].

Finally, Sha et al. [5] developed a 3D Finite Element Model to calculate the CDD in neighbouring tissues to the electrodes covered by a layer of conductive gel. Different conductivities were evaluated leading to the conclusion that the higher the gel resistivity, the greater the selectivity for electrical stimulation of the underlying tissues.

All the work briefly described above discussed the relevance of using mathematical modelling to predict the effects when electrically stimulating tissue.

This paper presents the development of 2D Finite Element Models used to assist the design of flexible printed circuit board based electrode arrays. The models

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were used to estimate the CDD beneath the tissue being stimulated by two elements of an electrode array, and they differ from those described above in their capabilities to simultaneously assess the influence of parameters such as electrode size, inter-electrode spacing and the conductive properties of the electrolyte in the CDD and its effect on the stimulation selectivity. Thereby, the models presented in this work allow the analysis of the existing trade off between current density and selectivity which is present at different array geometries and electrolytes of different conductivities.

Testing has been undertaken on two electrode arrays that are currently being developed as a part of a surface electrical stimulation FES system for drop foot correction following stroke and multiple sclerosis. A goal of this work is to be able to combine elements of the array to create 'virtual' and 'steerable' stimulating electrodes, controlled manually by the clinician/patient or automatically using feedback techniques.

Materials and Methods

Software: Two Finite Element Models were created using Maxwell 2D Student Version (Ansoft, Pittsburgh, USA). This interactive software uses generalized Finite Element based solvers to calculate problems regarding electric field and current density distributions (CDD) in structures with uniform cross sections. Thus, the AC conduction field simulator automatically solves for electric potential $\phi(x, y)$ using the following equation:

$$\nabla \cdot [\sigma E + j\omega\varepsilon\nabla\phi(x, y)] = 0 \qquad (1)$$

Where E is the electric field, σ the conductivity, ε the permittivity, $\phi(x, y)$ the electric potential at a given point in the plane x-y and ω the angular frequency at which the potential is oscillating. Once a model have been designed using the integrated development environment, values for materials conductivities, permittivities and boundaries are introduced by the user. The software then automatically generates the mesh on the model sketch by means of an adaptive algorithm which iterates until a preset threshold of error is reached.

Models: The models developed in this work replicate a cross sectional area for each electrode array. Figure 1 shows the sketch for both models.

The first array modelled was formed by twelve 3 mm wide elements of 50 micron height, equally interspaced by 3 mm. For the second array 2mm wide electrodes were used (same height) and the inter-electrode space was 4 mm. These parameters were chosen as a compromise between the required resolution of the array and the practicalities of fabrication. Two different and commercially available adhesive conductive gels were tested in both models as electrolyte for both arrays (Table 1).

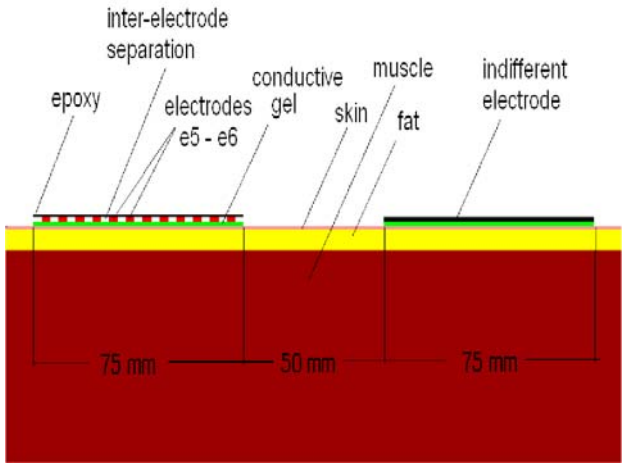

Figure 1. Model sketch designed using Maxwell 2D

Table 1: specifications for three different adhesive gels

Gel	σ (S/m)	Thickness (mm)
1	0.0033	0.89
2	0.066	0.89

Skin, fat and muscle were assumed as different layers of purely resistive materials with isotropic, homogeneous and time-invariant properties. Their conductivities were set according to values reported in the literature [9, 10]. Assuming that the stimulation waveform used in drop foot systems has a frequency between 20 and 50 Hz, the current density distribution in the tissue can be computed according to the quasi-static formulation [11]. Thus, assuming that permittivity can be set to be equal to zero in equation 1, then:

$$\nabla \cdot J = 0 \qquad (2)$$

$$J = -\sigma\nabla\phi(x, y) \qquad (3)$$

where $\nabla \cdot J$ represents the divergence of the current density. Each simulation was carried out at three different values of voltage (Vx = 80**,** 100 or 120V) that were applied to electrodes 5 and 6 of the array. Such electrodes were considered as equipotential sources of voltage. The indifferent electrode (7.5 cm long) was set to 0V. Finally, the normal of the current density vector at the external boundaries of the model was considered equal to zero.

Results

Table 2 lists the four simulations performed using different conductive gels and both models. Note that the number of triangular elements contained in each simulation mesh is enough to reduce the error, to around 0.12 %.

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Table 2: Simulations performed using Maxwell 2D

Simulation	Gel	Electrode Array	No. Elements	Error (%)
1	1	1	19848	0.1284
2	2	1	19848	0.1266
3	1	2	19847	0.1206
4	2	2	19872	0.1203

For each simulation, different coloured maps of CDD were obtained. The colour scale used for plotting the maps was constrained to a maximum value of 30 A/m^2 (equivalent to 3mA/cm^2), since this reflects reported values required to directly activate both sensory and motor neurons. Selectivity was defined as the spread of the maximum current density value towards both sides of the active group of electrodes (i.e. e5 and e6). Thus, the best pattern of selectivity will be the one that neither laterally extends further apart from the active group of electrodes, nor reaching any electrode at the vicinity.

Figure 2 illustrates the CDD when the array 1 is combined with the low conductivity gel (simulation No. 1). In figure 2.a, the current density under the electrodes e5-e6 and at the electrolyte-skin interface was evenly distributed at its maximum value (30.0 A/m^2). As the current penetrates into deeper tissue (i.e. muscle), the CDD fluctuates from 30.0 A/m^2 to 14.6 A/m^2, forming a triangular shape which narrows as the current travels towards the deepest muscle. Figure 2b presents a zoomed view of the electrodes e5 and e6 and their surrounding areas. It was observed that a current density of 30 A/m^2 extends to the sides, reaching the electrodes e4 and e7.

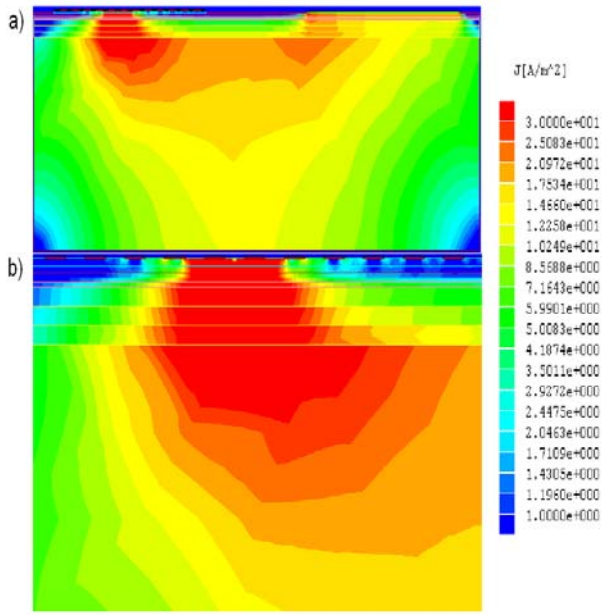

Figure 2. Simulation No 1, Vx=120V. a) normal view, b) close-up view of CDD around electrodes e5 and e6

In simulation No. 2, the array configuration remains the same, but the gel was replaced by the high conductivity one. Figure 3a shows a pattern of high current density which varies from 30.0 A/m^2 to 25.0 A/m^2 as the current travels towards the deepest tissue. However, figure 3b shows that the highest value for current density also spread out to the remaining electrodes forming part of the array.

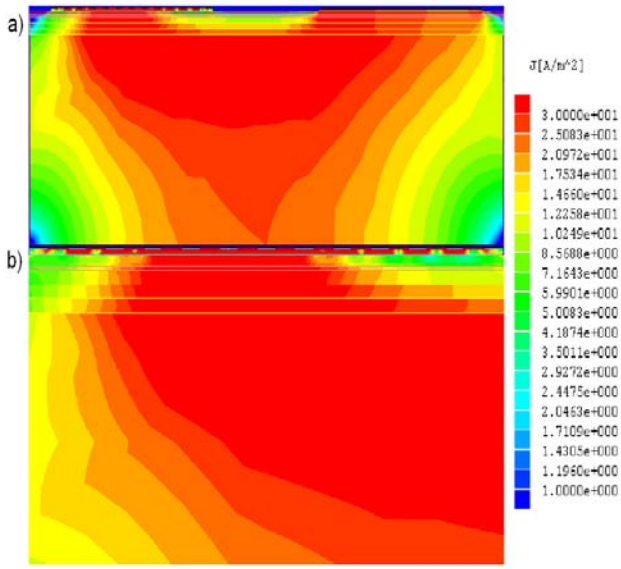

Figure 3. Simulation No 2, Vx=120V. a) normal view, b) close-up view of CDD around electrodes e5 and e6

Simulation No. 3 was carried out using the second array combined with the less conductive gel. Figure 4a shows a similar pattern of current density distribution immediately under the active electrodes and at the electrode-electrolyte-skin interface. Note that as the current travels down into deeper tissue, the CDD changes in similar fashion as reported for simulation 1, but within a different range of values (from 30.0 A/m^2 to 12.2 A/m^2). As shown in figure 4b, the current density also extends to the sides, being 20.9 A/m^2 the highest value reaching the electrodes at the vicinity.

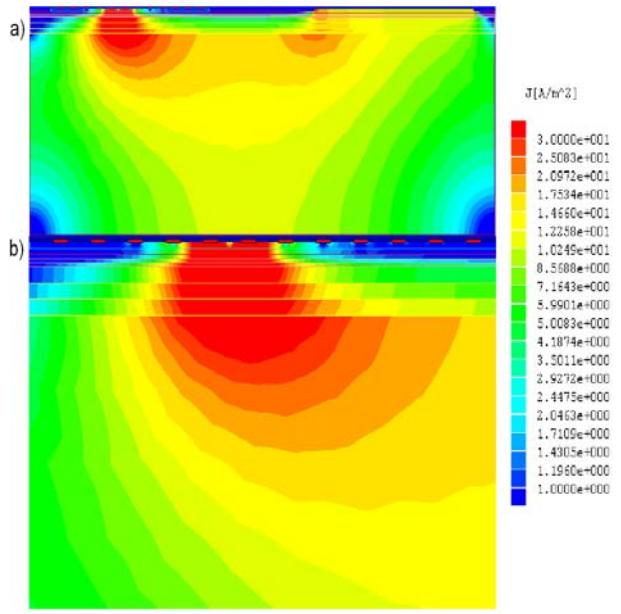

Figure 4. Simulation No 3, Vx=120V. a) normal view, b) close-up view of CDD around electrodes e5 and e6

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Finally, the second array combined with the high conductive gel was simulated. In figure 5a, a similar pattern of CDD to that reported for simulation 2 was obtained, but with a different range of values for current density (from 30.0 A/m^2 to 20.9 A/m^2). Figure 5b reveals once again how the current density at its maximum value spreads out from electrodes e5 and e6 to the rest of the electrodes on the array when using the high conductivity gel.

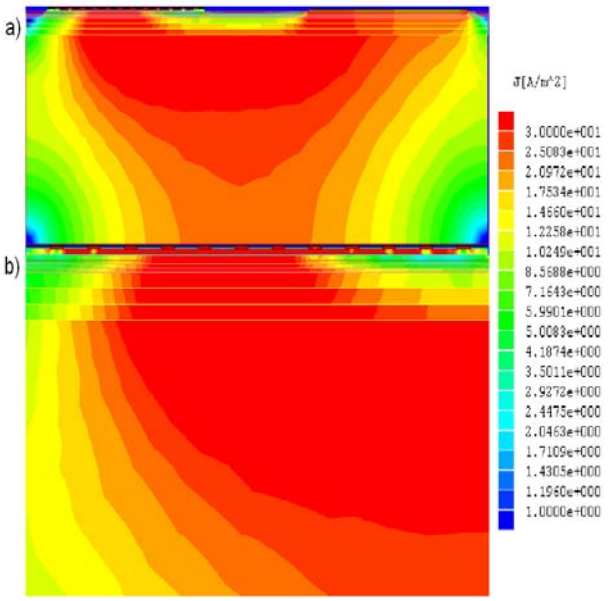

Figure 5. Simulation No 4, Vx=120V. a) normal view, b) close-up view of CDD around electrodes e5 and e6

Discussion

The effects of CDD on the selectivity underneath the stimulating electrodes have been previously reported in the literature as determinant factors for pain and effectiveness of electrical stimulation. Painful, hazardous and ineffective stimulation takes place when the current density is non-uniformly distributed at the electrolyte-skin interface and the underlying tissue [5, 7, 8].

Comparing different electrolytes, it was observed that both arrays allowed more current to pass into the tissue when they were used in combination with the high conductive gel (simulations 2 and 4). However, better selectivity was present when using the low conductive gel, even though the magnitude of CDD was reduced. Further comparisons were carried out between the two different arrays combined with the low conductive gel (simulations 1 and 3). The results revealed a improved pattern of selectivity for the second array, suggesting that this parameter can be improved when electrodes are smaller and further apart from each other. However, the current density is lower when using smaller electrodes separated by longer distances. Therefore, there is a compromise between the selectivity and current density which is related to the array geometry.

After modelling all the combinations of conductive gels and arrays, the results favoured the use of the second electrode array with gel No. 1 (simulation No. 3). It is believed that when using this configuration, the best compromise between CDD and selective stimulation can be achieved. The results obtained from this modelling are consistent with other studies [5,7], which demonstrated the use of low conductive electrolytes in order to minimize pain and maximise the effectiveness of electrical stimulation.

Conclusions

Parameters such as electrode size, inter-electrode spacing, and electrolyte conductivity and thickness play an important role in the effectiveness of electrical stimulation using surface electrode arrays. These parameters must be carefully chosen to provide an adequate functional response. Therefore, engineers must consider such parameters when designing surface electrode arrays in order to find the best compromise between current density distribution and selectivity. This work has proposed values for these variables. However, aspects such a time variant conductivities and non-uniform impedance of the skin were not considered during the simulations. Preliminary in-vivo experiments are being carried out in one normal subject using the array-gel combination proposed as the most suitable solution in this work. The initial results based on visual observations revealed that different patterns of foot response can be obtained while switching different groups of electrodes throughout the array, indicating good selectivity.

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