3D modelling of a hydrogel sheet - electrode array combination for surface functional electrical stimulation

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Abstract

This paper describes a 3D finite element model, which is being used to optimise the design of a self-adhesive electrode array for functional electrical stimulation (FES). The 3D model was built using ANSYS and used to calculate the current density distribution in tissues in the vicinity of the electrodes. The stimulating function, which is critical for nerve excitation, was then obtained. A series of hydrogel sheets with different resistivities, which were intended to ease application and removal of the electrode array, were tested in the model. The tissue volume with a stimulating function higher than the stimulation threshold of the target nerve formed a half-ellipsoidal pool. The 3D model is a useful tool to determine and optimise the spatial selectivity of the electrode array. The model demonstrated that for a 1mm thick hydrogel interface a minimum resistivity of 500Ωm is required to maintain selectivity.

Key words: FES, electrode array, hydrogel sheet, 3D modelling.

1 Introduction

Foot-drop is a common movement disability following stroke or other neurological pathologies [1]. A common approach to correcting it is to use functional electrical stimulation (FES). Current flowing from a relatively remote anode to a cathode placed below the head of fibula stimulates both the deep and superficial branches of the common peroneal nerve [2]. In order to produce optimal functional movement of muscles, it is necessary to correctly locate the electrodes, particularly the cathode, based upon the clinical knowledge of each individual patient. Difficulty in locating the correct electrode sites is one of the principal reasons for subjects discontinuing FES treatment [2].

We are therefore developing an electrode array for FES. Our approach is to build an array of ‘mini-electrodes’ with an overall size large enough to cover the possible stimulation site, allowing for errors in placement. The electrodes can be stimulated individually or in groups by an intelligent control system, which adjusts current to achieve the desired dorsiflexion and eversion of the foot [3]. In preliminary experiments we found that it was possible to ‘steer’ the response of the foot, using a prototype, non-adhesive array, but donning and doffing the array was found to be time-consuming [4]. It is postulated that an adhesive hydrogel sheet used between the electrode array and the skin may solve the donning problem, but the sheet will allow transverse currents to flow and so will tend to reduce the selectivity of the stimulating array.

The most relevant electrical parameter to achieve stimulation is the gradient of current density in the direction of the target nerve, often referred to as the stimulating function [1]. This can be modelled by a 3D finite element model to investigate the distribution of current under the electrodes and determine the optimal parameters of the electrode array. The influences of the electrical properties of the hydrogel sheet can be studied using such a model. Other parameters that can be modelled include the dimensions of the mini-electrodes and their grouping.

2 Methods

A 3D model was built using ANSYS (ANSYS Inc.). It had six components: cathode, anode, hydrogel sheet, skin, fat and muscle. The skin, fat and muscle were built using brick elements. The electrodes were located along the mid-line of the skin surface. The hydrogel sheet was placed between the cathode and the skin surface. The deep and superficial branches of the common peroneal nerve are the target nerves. They were assumed to be located 10mm...
below the skin surface and along its mid-line. The dimensions of their cross section were assumed to be negligibly small; hence they were not explicitly included in this model. There is a plane of symmetry through the centres of the electrodes and perpendicular to the skin surface. Hence only half of the model was required to represent the properties of the whole model. The dimensions and structure of this half-model are shown in figure 1. The origin of the coordinate system was placed at the bottom centre of the cathode.

The total input current was arbitrarily chosen to be 1mA. Due to the model symmetry, half of the current (0.5mA) was applied to the centre of the anode by setting the boundary condition of the node in the centre of the anode surface to be 0.5mA. The boundary condition of the node in the centre of the cathode surface was set to 0V. Because the current cannot flow through the boundaries of the model, all of the current delivered into the anode flows out from the cathode.

Compared with the linear dimensions of the mini-electrode, the human body was assumed to be infinitely large. To approximate to this in the model, it was therefore necessary to ensure that the model boundaries had a negligible influence on the current density distribution at the target nerve location. In order to check this condition, a test model with half of the dimensions on each boundary was built, as shown in figure 1. It was found that the maximum differences of the current density between the standard model and the test model were around 0.3%, which was deemed to be acceptable. It can be assumed that the difference between the standard model and the infinite model would be still smaller and hence this standard model can be assumed to represent an infinite model.

3 Results

The stimulation threshold of the target nerve is assumed to be an arbitrary value lower than the maximum stimulating function in a model. The model suggests that the stimulus pool, inside which the stimulating function was higher than the stimulation threshold of the target nerve, is half ellipsoidal (See figure 2). The target nerve will be excited if any part of it lies within the stimulus pool. However, the stimulus pool does not allow straightforward comparison of different parameters. Figure 3 shows the stimulus contours at the assumed target nerve depth of 10mm in the models without hydrogel, with conventional hydrogel (20Ωm), and with high resistivity hydrogel (500Ωm). It can be seen from figure 3 that the stimulus contour area reduces with increasing resistivity of the hydrogel sheet.

Due to the transverse current flow in the hydrogel sheet, the current in the model with hydrogel spread over a larger area than in the model without hydrogel. Hence the stimulating function was lower. An increased stimulation area may lead to inaccurate stimulation of target nerves and unnecessary pain caused by additionally excited pain fibres. In order to inhibit the transverse flow of the current, the thickness and the resistivity of the hydrogel
must be reduced and increased respectively. The thickness was chosen to be 1mm, the lowest value thought to be practical.

Higher resistivity of the hydrogel sheet leads to a higher voltage drop across it. The applied voltage must then be increased correspondingly. The effect of resistivity was investigated by modelling hydrogel sheets with different resistivities from 20Ωm to 1000Ωm. The magnitudes of the stimulus contour areas at a depth of 10mm for different hydrogel sheets are shown in Figure 4. It can be seen that the higher the resistivity, the greater the focality of the stimulation.

4 Discussion and Conclusions
Application of a self-adhesive electrode array to an FES system may ease its set-up and control. 3D modelling of the electrical stimulation is a useful tool that can be used to optimise the electrode array. By altering the parameters of the electrode array and the stimulator, the shape of the stimulus pool can be adjusted. A possible stimulus location on the target nerve can then be obtained. In order to maintain spatial selectivity for a 1mm thick hydrogel, the model predicts that the resistivity of the hydrogel should be 500Ωm or higher. For the hydrogel sheet with 500Ωm resistivity and 1mm thickness the voltage required to drive 1mA is 44V, 18% less than the required voltage in the model without hydrogel. This is due to the transverse current flow in the hydrogel sheet, which makes the effective size of the electrode appear larger. The required drive voltage increases above that required without hydrogel for resistivities greater than about 700Ωm.

As nerves were not explicitly included in this model, actual nerve stimulation was not directly investigated. The cathode in this model was only one mini-electrode rather than an electrode array. In the future, this model will be extended to include the nerve and an array of electrodes. The study of pain caused by electrical stimulation will be addressed by calculating the current density in the superficial layer.

References
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