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A finite element model to identify electrode influence on current distribution in the skin

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Abstract

Discomfort experienced during surface functional electrical stimulation (FES) is thought to be partly a result of localised high current density in the skin underneath the stimulating electrode. This paper describes a finite element (FE) model to predict skin current density distribution in the region of the electrode during stimulation and

its application to the identification of electrode properties that may act to reduce sensation. The FE model results showed that the peak current density was located in an area immediately under the stratum corneum, adjacent to a sweat duct. A simulation of surface FES via a high resistivity electrode showed a reduction in this peak current density, when compared with that with a low resistivity electrode.

Key words

Introduction

Functional electrical stimulation (FES), applied via surface electrodes, can be used to partially restore motor function which has been lost as a result of, for example, a stroke, spinal cord injury, or cerebral palsy [1,2]. However, FES is still used by relatively small numbers of patients. One aspect contributing to the low uptake is the discomfort experienced when current is passed through the skin [2]. Alternative approaches to reducing this discomfort have been widely examined, including varying the stimulus waveform and pulse width [3,4]. There is some evidence that changing electrode properties can affect the sensation [5-7] and this paper examines this area in detail.

The stimulation targets of surface FES are motor neurons. However, sensory receptors located in the skin lie between the electrodes and motor neurons, and also respond to

stimulation. For a long straight neuron axon, typically found in peripheral motor nerves, the gradient of electric field along the nerve (representing stimulation intensity) is probably the best measure of whether the nerve will respond to stimulation [8]. However, sensory receptors in the skin are convoluted and vary widely in their orientation with respect to the local electric field. Hence the sensory receptors will experience an electric field gradient along their length that is dependent both on the varying magnitude of the electric field and the orientation of the particular receptor within the field. As the orientation is unknown, the magnitude of the electric field, which is proportional to current density for a given medium resistivity, may reasonably be adopted as a measure of stimulation intensity for skin receptors [4]. A value of $0.1mA/cm^2$ has been cited as a representative estimate of the current density threshold for stimulation of most sensory receptors [4].

The skin is covered by the highly resistive stratum corneum (SC) lying on top of the epidermis and is traversed by skin appendages such as hair follicles and sweat glands. It is believed that stimulus current will preferentially flow through the skin via the low resistance skin appendages, acting as "current pores", possibly creating areas of high current density in their vicinity [9]. However, based on authors' knowledge, there has been little previous research to examine this effect in detail, or how the electrical properties of electrodes might influence the current distribution in the vicinity of skin appendages.

This paper describes the development of a FE model, which represents the anatomical structures in the skin and below, and the application of this model to explore the effect of a high impedance layer between the skin and the electrode. It was hypothesised that the introduction of such a layer would increase the impedances of all possible current pathways, reducing the effect of impedance differences between pathways and hence reducing peak current densities.

Methods

Prior to developing the model, a preliminary study was carried out to verify the hypothesis that a high impedance layer would reduce peak current densities. A pin array was constructed of discrete current paths, most of which were highly resistive, but with a single low resistance path. This experiment demonstrated that a high impedance layer between the current source and the pins would reduce the ratio of current through the low impedance path to that through the high impedance paths [10].

A 2D axi-symmetric FE model (see figure 1), centred on a current pore, was created, using a FE package (Ansys 10.0, Ansys Inc, USA). The skin, fat and muscle were modelled as horizontal layers and the skin was divided into SC and the rest of the skin (RS). The current flow in the vicinity of the current pore was assumed not to be influenced by the presence of any other current pores, and hence only one current pore (a sweat duct) was included in this FE model. As the average number of sweat ducts in the skin is about 1 per mm^2 [11], the width of the axi-symmetric model was selected

to be 0.5mm. The current pore traverses skin, fat and muscle. Representative values for the thickness of each component, together with the width of the current pore were taken from the literature [12,13]. Bone, blood vessels and sensory nerves were assumed to have negligible effect on the parameters of interest and were not explicitly included in the model. A metal foil stimulating electrode was modelled overlying a hydrogel layer on the skin surface. Differentiation between the materials in the model was achieved through assigning appropriate resistive properties to the elements representing the different media (see table 1). Apart from the SC, the other tissue properties are dominated by resistivity. To account for the capacitive properties of the SC, an equivalent resistivity was calculated at 1.67kHz (The pulse width was defined as 300μs for stimulation, and thus the equivalent frequency was approximately 1.67kHz.) to include both its resistive and capacitive properties. A convergence study was carried out to ensure the FE mesh-density was sufficiently fine.

As stimulation of a nerve for a given stimulation waveform is not a temporal summation but only varies with stimulation intensity, DC input current was used in the model. The applied input current was uniformly distributed over the surface of the foil electrode. Due to the limitation of the model size, the anode (common) electrode cannot be placed at a sufficiently remote position on the skin surface, and thus the bottom of the muscle was assumed to be at zero voltage and used as the grounding electrode. As current flow in the vicinity of the current pore was assumed not to be influenced by any other current pores, no current would enter or exit the model

laterally, i.e. via any outer surface of the model other than the foil electrode surface and the muscle bottom. Hence, the boundary conditions for the nodes on the model edges were defined as open-circuit to prevent current entering or exiting the model.

Results

As sensory receptors are normally located in the dermis, the current density distribution in the RS was assumed to determine sensation. The FE model predicted that the peak current density (hot spot) in the RS was located in the top corner of the RS layer, adjacent to the SC and the sweat pore (see figure 2). The non-uniformity of current distribution was quantified using a current hogging coefficient (CHC), which was defined as the ratio between the peak current density and the mean current density in the RS. The effects of four variables (hydrogel resistivity, hydrogel thickness, sweat duct resistivity, and SC thickness) on CHC were predicted by the FE model, as shown in figures 3, 4 and 5. Four different hydrogels were modelled, corresponding to commercially available samples (see tables 2 and 3), and their effects on CHC were plotted (figure 6).

Discussion

Figure 3 implies that using hydrogel with a higher resistivity leads to a markedly more uniform current distribution in the vicinity of the current pore. Varying hydrogel thicknesses from 0.3mm to 1.5mm has very little influence on current hogging. Therefore, hydrogel resistivity has the dominant effect on the current density

distribution for the range of hydrogel thicknesses modelled.

The resistivity of the sweat duct can be assumed to vary with the amount of sweat in the duct. Figure 4 suggests that current hogging peaks when the resistivity of the sweat duct is similar to that of the rest of the skin but the effect is not large. Hence, either sweaty skin or dry skin would be associated with somewhat more uniform current distribution in the RS during electrical stimulation.

Figure 5 shows that the thinner the SC the less current hogging occurs in the vicinity of a current pore. This suggests that reducing the thickness of the SC before stimulation will not only reduce the skin impedance, but also improve the uniformity of current distribution.

Four commercial hydrogels with a range of resistivities (see figure 6) were modelled. The results showed that current hogging reduced with increased hydrogel resistivity. The highest resistivity (hydrogel AG) resulted in a current hogging coefficient of 2.2, some 40 times lower than that found with the lowest resistivity hydrogel (hydrogel 703). This implies that the sensation associated with stimulation may be reduced by use of a high resistivity electrode.

The study limitations are as follows. Firstly, it has only examined current density in the vicinity of a current pore and does not consider other mechanisms by which high

current density can result, such as the electrode edge effect [4,7]. Secondly, no in-vivo validation of the model was conducted because of the difficulties of doing so without using unacceptably invasive methods. However the results of [6] tend to support the results presented here.

Conclusions

The FE model described here was used to predict the current density distribution in the skin during electrical stimulation due to the presence of a current pore. It can be used to predict the magnitude of the non-uniformity of the current density caused by anatomical parameters, such as SC thickness and sweat duct resistivity, and electrode parameters, such as hydrogel thickness and resistivity. The results suggest that a high resistivity stimulating electrode could reduce the discomfort associated with transcutaneous electrical stimulation, which is in agreement with our previously published experimental results [6].

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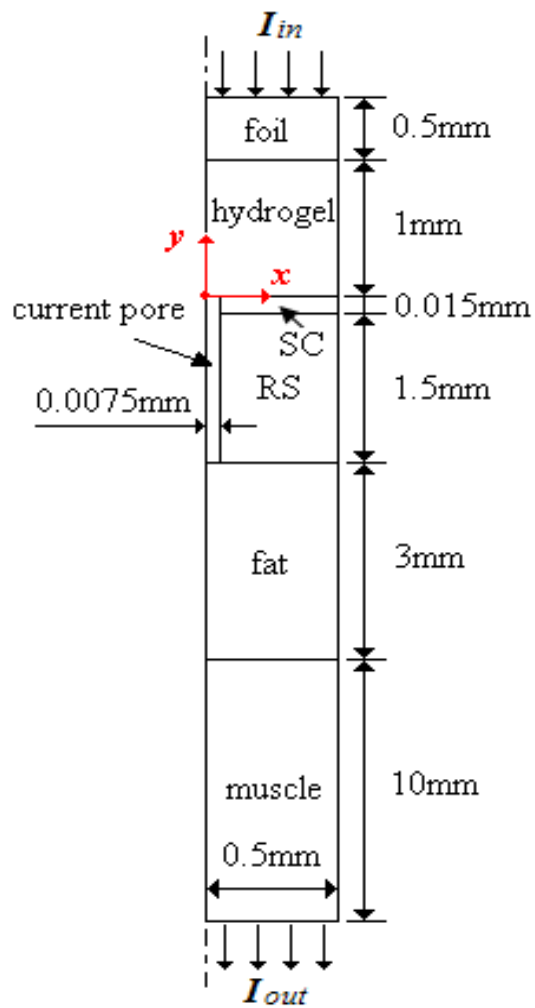
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Figure 1: Schematic of the FE model axi-symmetric around y axis,

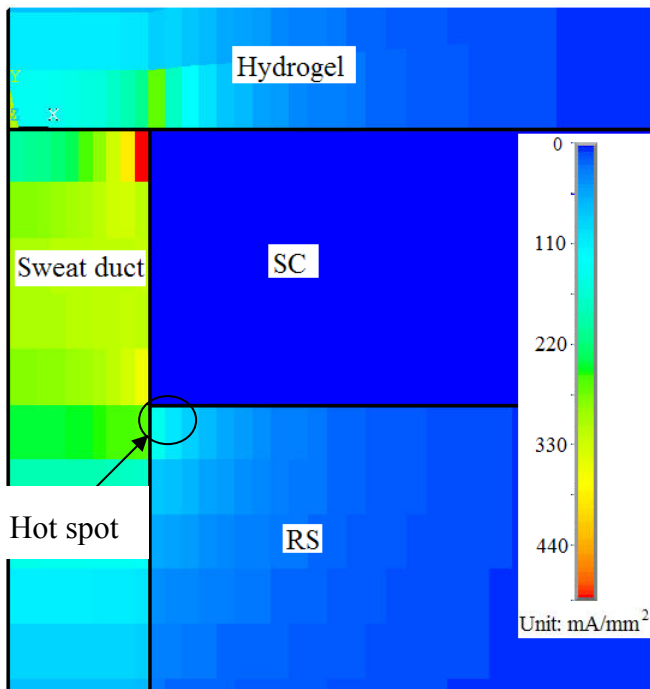indicating half of the cross section of a cylindrical model.

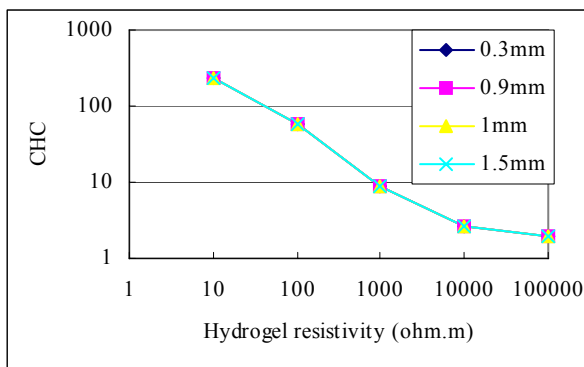Figure 2: Current density distribution showing the hot spot in the RS. Hydrogel resistivity: 55Ωm

Figure 3: Effect of hydrogel thickness and resistivity on CHC (all 4 curves are overlaid) SC: 0.015mm, Sweat: 1.4Ωm
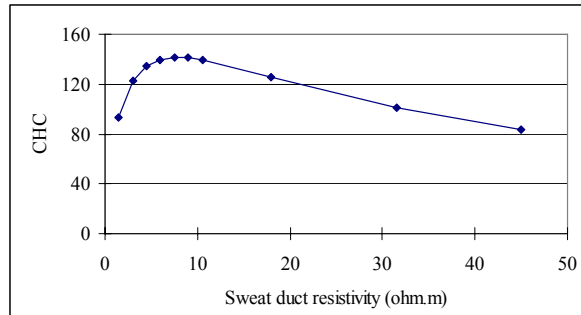
Figure 4: Effect of sweat duct resistivity on CHC

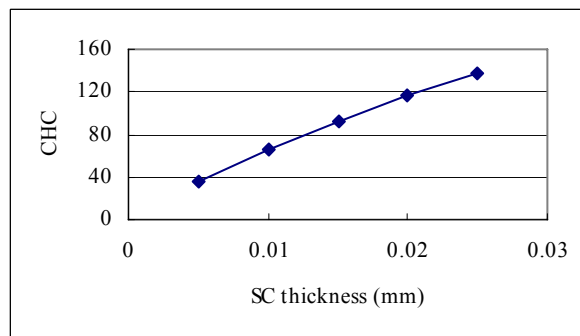SC: 0.015mm, Hydrogel: 100Ωm

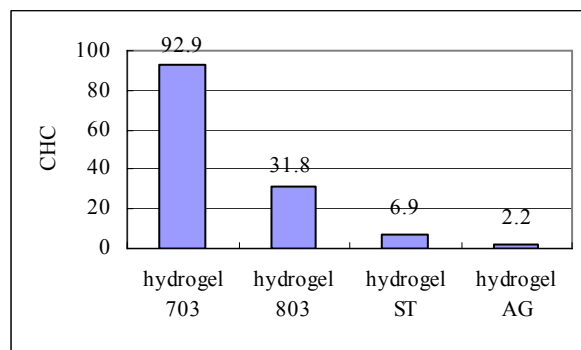Figure 5: Effect of SC thickness on CHC

Sweat: 1.4Ωm, Hydrogel: 100Ωm

Figure 6: Effect of hydrogel samples on CHC

SC: 0.015mm, Sweat: 1.4Ωm

Components	Resistivity (Ωm)
Foil electrode	1.5E-7
Hydrogel	$10-10^5$
SC	5300
RS	4.5
Fat	63
Muscle	2 on X, 4 on Y
Sweat	1.4-45

Table 1: Resistivities of the model components [10, 11]

Hydrogel name	Approx thickness (mm)	Resistivity (Ωm)
Hydrogel 703	0.9	55
Hydrogel 803	0.9	206
Hydrogel ST	0.5	1363
Hydrogel AG	0.3	25185

Table 2: Properties of the hydrogel samples

Hydrogel name	Product code	Suppliers
Hydrogel 703	AG703	Axelgaard manufacture Co., Ltd. USA
Hydrogel 803	AG803	Axelgaard manufacture Co., Ltd. USA
Hydrogel ST	SRBZAB-05SB	Sekisui Plastics, Co., Ltd. Japan
Hydrogel AG	AG3AM03M-P10W05	Sekisui Plastics, Co., Ltd. Japan

Table 3: Product details of the selected hydrogels