

Motor Cortex Stimulation: a computer modelling study

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Abstract

In this paper initial results obtained from computer modelling of Motor Cortex Stimulation for chronic pain treatment are presented. The model predicts that the thickness of the cerebrospinal fluid under the epidural electrode has a great influence on the electrical field in the motor cortex and therefore on fibre stimulation thresholds. It is further concluded that afferent fibres are easier to excite than efferent fibres of the same diameter, implying that their stimulation may be responsible for the analgesic effects. However, additional anatomical data and improvements of the volume conductor and fibre model are needed to allow a better prediction which neural structure responds best to a specific stimulation. Ultimately, this modelling study should help to understand the mechanisms of motor cortex stimulation and help to improve its efficacy.

1 Introduction

Motor Cortex Stimulation (MCS) is a developing clinical technique used to treat certain chronic pain syndromes. It was introduced by Tsubokawa et al. for poststroke central pain [1] and by Meyerson et al. for trigeminal/ facial pain [2], but may possibly be used for other indications as well [3][4]. The technique involves the insertion of a quadripolar plate electrode epidurally over the motor cortex (precentral gyrus) followed by the electrical stimulation at 50% or less of the motor threshold which results in the analgesic effect [4]. However, since the exact biophysical mechanisms underlying this process are not yet known, different electrode positions, contact combinations and stimulation parameters are used with varying success. Understanding the mechanisms would help standardise the technique and improve its efficacy. As with Spinal Cord Stimulation (SCS), computer modelling may be an invaluable tool to achieve these goals.

2 Methods

2.1 Motor Cortex model

UTSCS software [5] was used to develop a 3D volume conductor model of the human motor cortex. The model comprises the precentral gyrus with surrounding sulci, white matter, cerebrospinal fluid (CSF), the skull and the stimulating electrode as shown in Figure 1. The dimensions and electrical conductivities of different model compartments were set to match empirical data. The thickness of the CSF underneath the electrode was set at 1.1 mm.

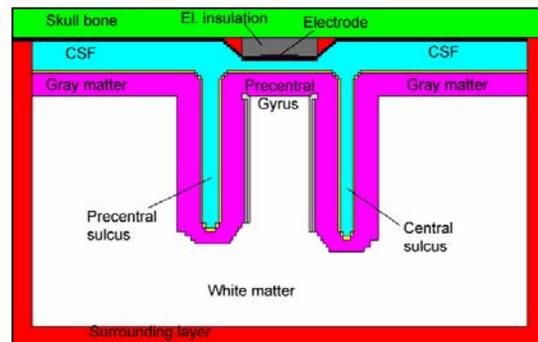


Figure 1: A transversal cross-section of our motor cortex model.

The electrical conductivity of the artificial boundary layer was adjusted to match empirical impedance data. Monopolar (cathodal) stimulation with rectangular 210 μ s voltage pulses was simulated. The electrical field in the volume conductor imposed by the stimulation was calculated using the finite difference and red-black Gauss-Seidel numerical methods.

2.2 Nerve fibre models

In order to calculate the response of the neural structures in the motor cortex to the imposed electrical field, a McNeal type based model of myelinated human fibres was used [6]. Three fibre types were modelled:

- 1) 'A1' (Figure 2A) – horizontal part of a thalamo-cortical fibre (afferent)
- 2) 'A2' (Figure 2B) – thalamo-cortical fibre (afferent): main fibre entering the cortex

perpendicularly and bifurcating parallel to its surface as modelled in type A1 and

3) 'E1, E2 & E3' (Figure 2C).– cortico-thalamic and cortico-spinal fibres (efferent).

The diameter of the fibres was varied from 5 to 15 μm , representing the range in which our fibre model has been validated [6] (and not necessarily the range of fibre diameters present in human motor cortex). The depth of the horizontal part of the A-fibres in the gray matter was varied from 0.7 to 1.6 mm. The depth of the initial horizontal part of the E3-fibre was varied from 3 to 4.6 mm below the upper surface of the motor cortex, and its proximal end at 1.4 mm in the cortex (Figure 2C).

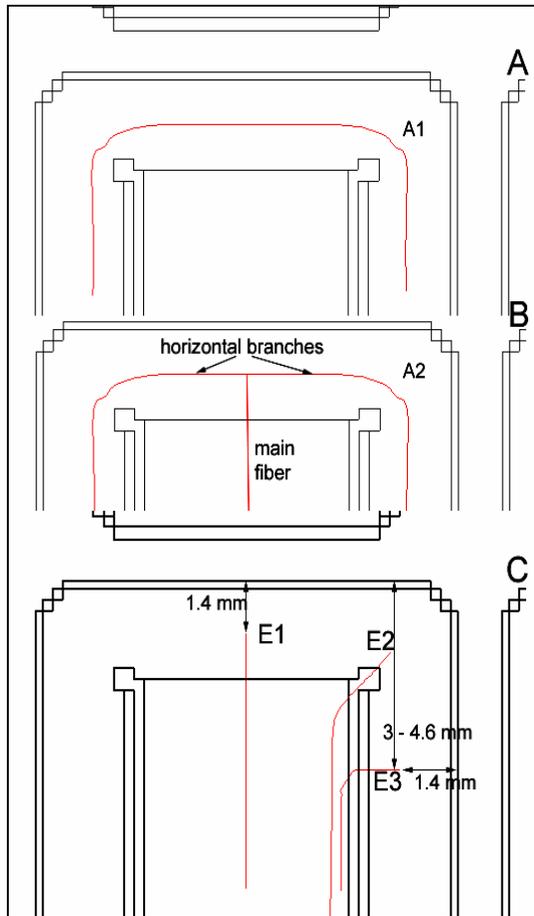


Figure 2: Modelled fibres in the motor cortex. A & B) Thalamo-cortical afferents C) Cortico-thalamic and cortico-spinal efferents.

The output parameter of the simulations was the voltage to be applied in order to excite a particular fibre – ‘threshold voltage’.

3 Results

3.1 Electrical field

The stimulating electrode imposes an electrical field in the surrounding tissues. We found that the thickness of the CSF under the electrode has an immense influence on this field, whereas the width of the sulci does not. A reduction of the CSF thickness results in deeper penetration of the field in the cortex with same voltage applied. The iso-current density lines in a cross-sectional plane through the electrode are shown in Figure 3.

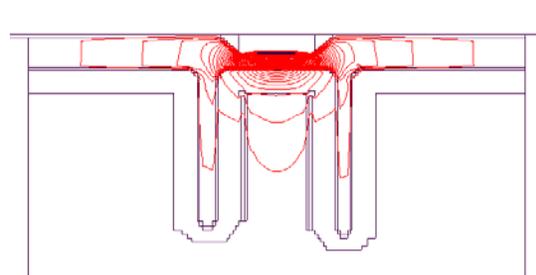


Figure 3: Iso-current density lines near the stimulating monopolar contact. 100 lines equally spaced in the range 1.4 – 144 $\mu\text{A}/\text{mm}^2$.

3.2 Threshold voltages

The threshold voltages of the fibres decrease as the diameter of the fibres is increased. Having the same fibre diameter, the A2 fibre has the lowest threshold, followed by the A1, E3 and finally E2 fibre having the highest threshold (Figure 4). The E1 fibre, being perpendicular to the cathode, is hyperpolarized at any given voltage and therefore cannot be excited.

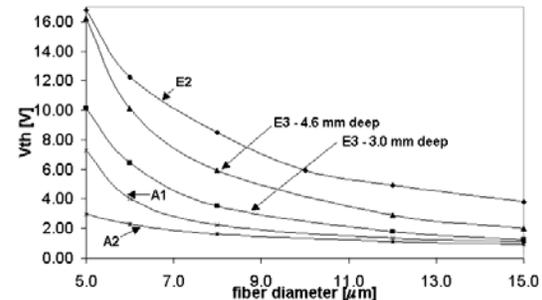


Figure 4: Excitation threshold as a function of fibre diameter. A-fibres have a horizontal part at a fixed cortical depth (1.4 mm); E3 fibre thresholds presented are the extremes.

With an increasing depth of the fibre (see Methods), the threshold of the fibre is increased. An example, using 8 μm diameter fibres and various depths in the motor cortex is shown in Figure 5.

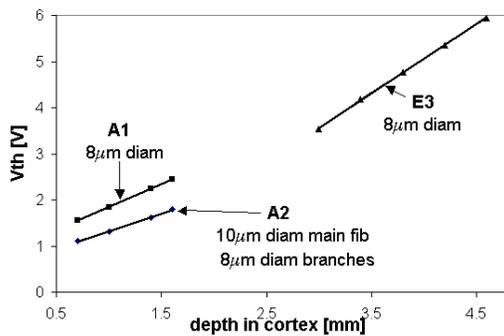


Figure 5: Fibre threshold as a function of cortical depth. A-fibre depth is measured from the cortical surface to the horizontal branch of the fibre. E3-fibre depth is measured from the upper cortical surface to its horizontal part running perpendicular to the sulcus (see Figure 2).

As shown in Figure 5, the relation between fibre threshold and its depth is fairly linear for all modelled fibres.

While the diameter of the horizontal branches of the A2 type fibre model was kept constant at 8 μm and its depth at 1.4 mm in the cortex, we varied the diameter of the main fibre from 5 to 15 μm . The threshold decreased from 2.2 to 1.3V.

4 Discussion and Conclusions

We developed a model for Motor Cortex Stimulation using the same tool we have used previously to model the effects of Spinal Cord Stimulation - UTSCS software.

4.3 Volume conductor model

The modelling results show that the thickness of the CSF layer underneath the electrode is an extremely important parameter. The thickness of this layer is not yet known. It may change after implantation and may vary among subjects. The electrical conductivities of some tissues are not well known and we had to use approximations to match empirical impedance values.

4.4 Nerve fibre models

1) Afferent fibres get more easily excited than efferent fibres of the same diameter at any modelled position. However, there is evidence that the diameter distribution of efferent fibres is centered around higher values than the distribution of afferent fibres, implying that thresholds of smaller A-fibres should be compared with those of larger E-fibres. The exact diameters and exact fibre orientations and positions are though not (yet) known, but they

are needed to allow a reliable prediction of the response of the neural structure.

2) The behaviour of our fibre model has been validated in the range 5-15 μm [6], which may not be a sufficient range to model motor cortex stimulation.

3) The proximal end of our fibre models is assumed to be sealed with an infinite impedance. This may not be good enough for the E-fibre type that is terminated by a soma. Although the input impedance of a soma is rather high, this seal may have influenced our predictions of the E-fibre threshold.

4) As shown, modelling a branching point in an A-fibre reduced the fibre threshold differently for different diameter ratios. However, this ratio and the branching point position are also not known.

4.5 Conclusion

This study is an early step in modelling the effects of Motor Cortex Stimulation which should (with additional improvements) allow to identify the neural elements stimulated in the therapy. This would result in the understanding of the immediate effects of stimulation, improved electrode and stimulation strategy design and ultimately improvement of the therapeutic efficacy.

References

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