The 3D potential induced by functional electrical stimulation with multi-contact cuff electrodes: simulation and validation.

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Abstract

Our aim is to propose a numerical model of nerve-cuff electrode which will be used to study and predict interactions between nerve fibres and electrode during a FES. The contributions concern both the numerical resolution of the problem and the experimental validation that is currently carried out using a robotized probe and an electrode designed specifically for the experiment. The paper describes an accurate way to compute current densities and voltage within the nerve which can be used to determine whether an axon is fired or not depending on its position.

1 Introduction

Functional Electrical Stimulation (FES) provides a way to restore movement of paralyzed limbs by activating efferent somatic axons. Indeed they innervate the striated muscle fibres that can thus generate force when stimulated. FES can also be used to activate other target organs such as the detrusor smooth muscle for bladder control. The principle remains the same i.e. firing motor axons of the desired target muscles. However, within a nerve composed of different types of axons innervating muscle fibres, an efficient implanted FES system must meet the challenge of selectively activating these fibres; for instance, for bladder control, one must be able to stimulate the detrusor (smooth muscle) without stimulating the striated sphincter (innervated by the same sacral root); for movement restoration, to limit fatigue effect, the slow fibres should be stimulated instead of the fast fibres of striated. Empirical tests [1, 2, 3] show that fibre selectivity is possible depending on the geometry of the electrodes and the shape of the stimulus. Some studies focus on the interaction between nerve fibres and the stimulating electrodes. Among these, [4] examines the behaviour of axons excited by electrical fields. In [5], simulations show that the diameter dependency of nerve fibre recruitment is influenced by the electrode geometry while [6] predicts that the intraneural or intrafascicular electrodes are necessary for selective stimulation of non-superficial nerve fascicles. Authors in [7] conclude that a transverse tripoles activates superficial nerve fibres in a more selective way than other configurations (monopolar, bipolar,...). Using a volume conductor, [8] determines a minimal quantity of charge per pulse needed for selective nerve stimulation depending on pulse shapes.

Our aim is to propose a numerical model of nerve-cuff electrode which will be used to study and predict the electrode/nerve interaction between fibres during a FES. This model will be used to determine the optimal geometrical and electrical parameters for a multipolar cuff electrode for selective neural stimulation. This kind of study has been investigated in [9], but only considering a 2-D conductor model. Here we consider a 3-D problem and use a Boundary Elements Method recently proposed for EEG, to compute the electrical potential due to an applied boundary current [10,11].

Our physical model, of which preliminary results have been presented in [15], is based on the quasi-static approximation of Maxwell’s equations. We describe the BEM, show simulation results, and describe an experimental setup which is being used to validate the numerical model.

2 Methods

2.1 Physical model

The electrical potential $V$ generated by a cuff electrode around a conductor satisfies the Poisson equation, deriving from the quasi-static Maxwell equations: $\nabla \cdot (\sigma \nabla V) = 0$ within the nerve volume, and $\sigma \partial V/\partial n = j$, on the surface, where $j (A)$ is the applied current.

Fig. 1 Nerve model consisting of two nested cylinders, and cuff electrode with three four-contact rings.
The stimulated nerve (Fig. 1) is modelled as two nested cylinders, of lengths 80 mm and radii 1 and 1.25 mm. The inner cylinder models the fibre bundles ($\sigma_1 = 0.6 \text{S/m}$), and is surrounded by connective tissue ($\sigma_2 = 1.7 \text{S/m}$). The nerve is supposed to lie in a non-conductive medium ($\sigma_{\text{exterior}} = 0$), and the only current crossing the nerve surface is at the electrode contacts. A stimulation current is applied through a 12-contact cuff electrode composed of three rings. The applied current $j$ must follow $j = \int \Phi_i(p) \sigma_{ij} ds(p)$. The width of the rings along the $z$ axis is denoted $a$. The distances between the two rings are $d_1 = d_2 = 5 \text{mm}$. The angular widths and distances of the contacts are $\alpha = \pi/10$ and $\beta = 4\pi/10$ (Fig. 2).

We have extended this BEM to also compute the potential at any position $p$ within the inner nerve cylinder, using the integral representation:

$$V(p) = \sum_i \left( \frac{\| p - p_T \|}{4\pi} \int_{T_i} \frac{\Phi_i(p)}{\| p - p' \|^2} ds(p') \right) v_i + \frac{1}{4\pi} \sum_i \int_{T_i} \frac{ds(p')}{\| p - p' \|^2} (\sigma \Phi_i)_{ij},$$

where $T_i$ are triangles of $S_1$, $p'$ points of $T_i$, $p_T$ the barycentre of $T_i$ and $\Phi_i$ is the piecewise linear finite element associated to $T_i$.

## 2.3 Experimental setup

In order to validate the numerical simulation, a four times scaled phantom nerve has been built up, consisting of a plastic tube filled with saline solution surrounded by 12 metallic contacts according to the model of Fig. 1. Constant current synchronous electrical stimulations are applied on different electrode configurations while a micro robotic arm sweeps in 3D and records potentials within the setup. Stored data are post processed.

![Fig. 3 A robot controls the 3D position of the sensing electrode measuring the potential induced by electrical stimulation within the phantom nerve.](image-url)

## 3 Results

An example of numerical simulations performed with the OpenMEEG software package [16] is shown in Fig. 4. The outer surface mesh is displayed in the bottom picture, and the electrode rings are represented as a change of color in the mesh. For a current applied to two opposing contacts of the center ring, the upper left (resp. right) picture shows the potential at $z=0$ (resp. $x=0$).
Fig. 4 Example of numerical simulation: on the planes \( z=0, x=0 \) (top) and on the two planes simultaneously (bottom), for a current applied to two opposing contacts of the middle electrode ring (centered longitudinally at \( z=0 \)).

Experimental measurements are made at MXM plant in order to validate the numerical simulations. Preliminary results are displayed in Fig. 5, for a cubic container (left) and a cylindrical phantom (right). The results of the cylindrical phantom match the simulation results of Fig. 5.

This work, in progress, will allow to optimize the parameters of the numerical model, in terms of number of mesh elements, or length of cylinders to be considered to validate the numerical model.

Fig. 5 Measured data, for two different geometries: on a cubic container (left) and on a cylindrical phantom electrode (right).

4 Discussion and conclusion

The computational approach described in this paper, provides a efficient compromise between accuracy and computational. It was initially developed for magneto-electroencephalography (MEEG), where it has demonstrated its excellent accuracy. The proposed method allows to compute a potential in the 3D volume without the need for complicated 3D meshing. Only the interfaces between tissues of different conductivities need to be meshed, and the 3D potential derives from Boundary Integral equations.

The results presented here are a preliminary assessment of the validity of the simulated 3D potential in FES. They are to be conducted thoroughly, in vitro as presented above, as well as in vivo.

Accurate numerical simulation of the 3D potential within a nerve leads the way to optimization of stimulation parameters, both in the geometrical configuration of electrode, and in applied current patterns. The use this forward nerve model to investigate this optimization issue may lead to minimizing charge injection while maximizing selectivity abilities.

References

[16] OpenMEEG software: www-sop.inria.fr/odyssee/software/OpenMEEG