31st Annual International Conference of the IEEE EMBS
Minneapolis, Minnesota, USA, September 2-6, 2009

A toolchain to simulate and investigate selective stimulation strategies for FES

Jeremy Laforet, David Guiraud, *Member, IEEE*, and Maureen Clerc

***Abstract*— When contracting a muscle using NFES (Neural Functional Electrical Stimulation), the stimulus always activates the axons of greater diameter first. Also selective activation of given fascicle inside a nerve is not possible with classical cuff electrode as the recruitment is performed uniformly around the nerve. These limits lead to poorly selective muscle recruitment, inducing fatigue and possible pain. To overcome this, selective stimulation strategies can be used.**

We propose a toolchain to investigate, simulate and tune selective stimulation strategies. It consists of a conduction volume model to compute the electric field generated in the nerve by a cuff electrode surrounding it; an axon model to predict the effect of the field on the nerve fibre — the generation, propagation and possible block of action potentials; and an interface script that links the two models and generates the code of the input function for the nerve fibre model.

We present some simulation results to illustrate the possibilities of the toolchain to simulate such strategies. Ongoing experimental validations are also discussed. They will enable us to tune the model and may lead to further improvements.

***Index Terms*— multipolar electrode, simulation, Functional Electrical Stimulation**

I. Introduction

To contract a muscle, the central nervous system (CNS) activates motoneurons, which send neural commands —action potentials— to the muscle fibres. Each of them controls a definite set of homogeneous muscle fibres, a motor unit. Thus, the recruitment of a muscle consists in motor unit selection, depending on the number of fired motoneurons. Moreover, CNS selects the type of fibres to activate depending on the task to achieve: fast twitch fatigable (IIa, IIb) or slow twitch fatigue resistant (I) ones.

The CNS activates a muscle gradually, starting with type I fibres, then adding the type II fibres if more force is needed. It also activates different sets of motor units over time to limit fatigue.

When contracting a muscle using NFES (Neural Functional Electrical Stimulation), the stimulus always activates the axons of greater diameter first. Then, as the amplitude is raised, it will recruit the smaller ones. Unfortunately, these big fibres correspond to type II and the smaller ones to type I. Besides, classical NFES stimulus shapes evoke both efferent and afferent action potentials so that unwanted reflexes may occur, or, in some pathologies, pain can not be avoided.

J. Laforet is with DEMAR project team, LIRMM, University of Montpellier 2, 161 rue Ada, Montpellier, France laforet@lirmm.fr

D. Guiraud is with the DEMAR project team, LIRMM, INRIA, 161 rue Ada, Montpellier, France guiraud@lirmm.fr

M. Clerc is with Odyssée project team, INRIA, 2004 Route des Lucioles, Sophia Antipolis, France Maureen.Clerc@sophia.inria.fr

Finally, NFES recruitment with classical cuff electrodes is performed uniformly around the nerve making difficult to activate fascicles or different parts of the same muscles selectively.

In order to achieve inverse recruitment order, afferent-efferent selectivity and spatial localisation of the focus stimulus point within the nerve, multipolar electrodes and advanced stimulators have to be used. We developed both to experimentally investigate and assess these possibilities furthermore described in literature. Many are being investigated and they can be divided in two groups: geometrically selective methods and diameter selective ones.

The aim of the first group is to activate independently fibres from different fascicles — group of nerve fibres that will later branch out. The poles of a multipolar electrode around the nerve (Fig. 4) can stimulate different sectors of it. However selectivity is limited —in the example the smallest zone which can be stimulated is roughly a quarter of the nerve. After a surgery, the angular position of the cuff relatively to the nerve remains unknown.

An evolution of this solution is to use non-cylindrical cuff to increase the ratio between nerve/electrode contact surface over cross-section surface. Using flat cuffs [1], it is then possible to reshape the nerve slightly to achieve better selectivity.

Intrafascicular electrodes [2] may be highly selective: only few nerve fibres around the electrode are stimulated but it is not possible to activate large fascicles or several smaller ones with a single electrode. The method is too invasive to be used on humans.

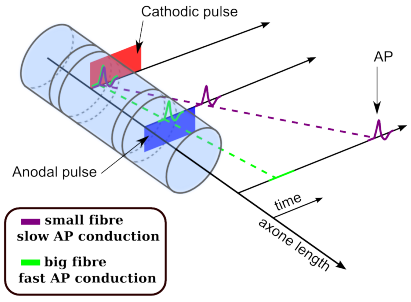

Fig. 1. Principle of anodal block.

The second group of methods aims at reproducing the natural order of recruitment of the neural fibres. A way to achieve this is the anodal block [3]. It uses the hyperpolarisation that can occur under the anode to stop action potentials. As bigger fibres have a lower stimulation threshold, they

are fired first. Increasing stimulation intensity a few tens of milliseconds later hyperpolarizes the membranes under the anode and then blocks these action potentials while smaller diameter fibres are activated under the cathode (Fig.1). Therefore, it needs longer stimulations with accurate timing to be efficient.

Another strategy is to add a high frequency (a few hundreds Hz) to the usual stimulus (a few tens of Hz) [4]. This new component of the stimulation will inhibit the conduction of some action potentials. The amplitudes can be tuned separately to select which fibres have to be activated or blocked. The main drawbacks are the great amount of charge injections and the unknown physiological phenomenon that explains the principle.

We already developed multipolar electrodes and benchtop stimulators with industrial partners to provide for future implantable stimulation units, with up to 12 poles, and arbitrary stimulation shape [5]. Even though literature describes the principles of the strategies to be used, tuning remains a tricky problem. This is the reason why we develop a modelling toolchain, not only to simulate and investigate selective stimulation, but also to optimise it, mainly as regards charge injection minimisation, and selectivity level. To achieve this, we present the first simulation results of an original method that computes 3D current flows as well as spike generation using the Neuron software.

II. Inside the electrodes

We concentrate on cylindrical cuff electrodes as they are widely used and easy to manufacture. We model the contact between the electrode and the nerve, and we simulate the electric field generated inside the nerve.

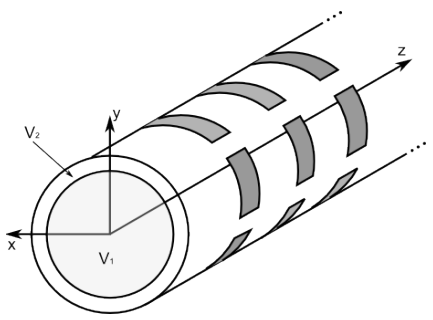

Fig. 2. Two cylinders nerve model and cuff electrode.

For this purpose OpenMEEG was used and adapted [6]. It provides a 3D conduction model of the electrode and nerve. They are represented as coaxial cylinders of non isotropic but homogeneous conductivities — the nerve itself and its envelope of connective tissues (V_1 and V_2 in Fig. 2). We consider a 12-contact cuff, and we take as input the signed current applied on each one.

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

The surfaces are represented with triangular meshes and the Poisson equation is solved using a Boundary Element Method known as the Symmetric BEM. Its outputs are the potential field computed at the vertex of each triangle defining the separation surfaces and the current density normal to the triangle.

The method accuracy and convergence rate are higher than classical finite element methods thanks to the limited computation performed on the boundaries only [7]. It is furthermore possible to compute the potential and the current density vector at any position within the inner nerve cylinder volume to provide the data needed as input to the Neuron software [8].

Using the integral representation, the potential at position p is given by :

$$V(p) = \frac{1}{4\pi} \sum_j \left(\sum_{i|j \in T_i} \int_{T_i} \frac{\Phi_j(p')(p-p\prime) \cdot n}{\|p-p\prime\|^3} ds(p\prime) \right) v_j + \frac{1}{4\pi} \sum_i \int_{T_i} \left(\frac{ds(p\prime)}{\|p-p\prime\|} \right) (\partial_n v)_i, \quad (1)$$

where j indexes the vertices of the mesh, the T_i are triangles, $p\prime$ points of T_i, n the normal to triangle T_i, and Φ_j is the piecewise linear finite element associated to node j on triangle T_i.

These values are computed on a cubic grid, designed so that its nodes match the positions of the Ranvier nodes of the nerve fibres we aim to simulate.

III. From electric fields to action potentials

We use the simulated electric fields to predict the activation of nerve fibres depending on their characteristics (mainly the diameter) and their position within the nerve (given by the position of the grid used to estimate potentials). The fibre model was designed using Neuron [9]. It has to take into account the excitation of the axon's membrane by the electric field in order to compute the propagation of the generated action potential. The model we defined, is able to simulate axons of various diameters so that it exhibits their different behaviours under the same stimulation profile.

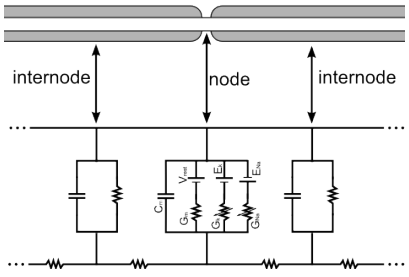

Fig. 3. Electrical circuit for a myelinated axon

The myelin sheet speeds up conduction of action potential and isolates electrically the axon, but induces discontinuities. In between myelinated zones lie the Ranvier nodes, where the

TABLE I

PARAMETERS OF THE CONDUCTION MODEL FOR A SMALL MAMMALIAN NERVE.

Parameter	Value
Contact size	0.5mm
d1	5mm
d2	5mm
Inner nerve length	40mm
Inner nerve diameter	1.3mm
Inner nerve conductivity	0.6S/m
Outer nerve length	40mm
Outer nerve diameter	1.5mm
Outer nerve conductivity	1.7S/m

action potential may arise. As we consider only myelinated fibres, we have to know the value of the field only at the Ranvier nodes. Thus we apply the value of the electric potential field to the exterior of the axon's membrane at each node. If the induced membrane depolarisation is great enough an action potential will be generated. Then, the action potential will travel in both directions along the axon.

To model the axon we use two types of membrane models as in [10]. For the myelinated section we use a simple passive model and for the nodes we use a Hodgkin-Huxley model. This model describe the generation and propagation of action potentials. It relies on the ionic currents flowing throught of the membrane. The model is illustrated in figure 3. We consider whole fiber diameters, including the myelin thickness.

The interface between OpenMEEG and this model is a Python script that takes the values of the electric field from files and generates the code of the axon model input function. It generates the spatiotemporal waveform of the stimulus and applies it at the Ranvier nodes. To do so, it creates variable tension sources on the outer side of the axon membrane at the node positions.

The inner membrane potential in each node of the axon is collected at every time increment and saved in a file as a matrix. This gives the evolution of the membrane potential along the axon, during the generation and propagation of the action potential.

IV. SIMULATION: MULTIPOLAR STIMULATION

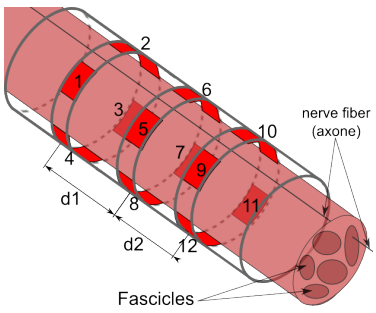

Fig. 4. Configuration of the simulation example.

To illustrate the possibilities of this toolchain we model a small mammalian nerve and the activation of one axon. We

TABLE II

PARAMETERS OF THE FIBRE MODEL.

Parameter	Value
Fibre diameter	10μm
Internode distance	1150μm
Number of nodes	60
Temperature	37°C

simulate tripolar stimulation with a monophasic rectangular pulse of 500μs duration. Contacts 5 to 8 behave as a ring cathode while the eight others behave as two ring anodes. For these simulations all parameters were taken from the literature.

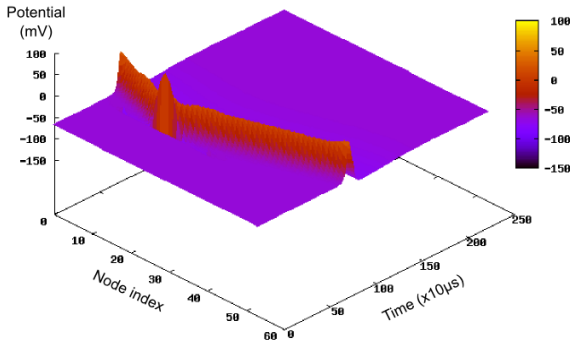

Fig. 5. Plot of the model output: inner membrane potential. Anodes are located just before node 10 and 30, the cathode is near node 20. The view point lets see only cathodic artifact happening at 0.5ms.

We simulate a part of the axon long enough to see the propagation of the possible action potential outside of the generated field. To do so, we settled for a length including 60 nodes. Fig 5 is a 3D plot of the output of the model. The elevation of the plot is the membrane potential. It is presented as a function of both time and position along the axon (by the index of the node). This view enables to see the whole propagation and possible blocking phenomena on a single plot.

The artifact happening at t=500μs around node number 20 is due to the onset of stimulation at the cathode level. Similar artifacts but of opposite sign are present near the anodes; they cannot be seen due to the viewpoint of the graph.

The shape of the action potential is shown on figure 6. It presents the value of the membrane potential at node number 40 which is located outside the electrode cuff. The graph shows the characteristic phases of the action potential. The conduction velocity computed is around 70m/s; it corresponds to the values in the literature for a myelinated axon of this diameter.

To determine the threshold diameters below which no stimulation occurs, above which the axon is fired, but also eventually blocked under the anode, we play with the axon parameters, with the same set of data issued from OpenMEEG. To provide the spatial mapping of activation, only the

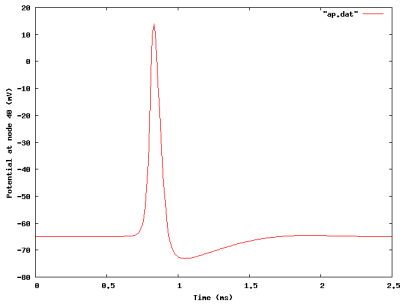

Fig. 6. Membrane potential at node 40.

position of the axon within the nerve has to be changed, i.e. the computation of axial potentials using the same Neuron model and the same OpenMEEG simulation (only some potentials within the homogeneous volume are calculated),

V. Validation

To validate simulation made with OpenMEEG, measures of the actual field generated in a physical model of cuff is currently performed. This will enable to tune the parameters of the electrode model. This setup is build around a four times scale nerve phantom. It is a plastic tube filled with saline solution and surrounded by 12 metallic contacts as in the model shown figure 2. The potential will be recorded by a probe whose positioning is computer-guided.

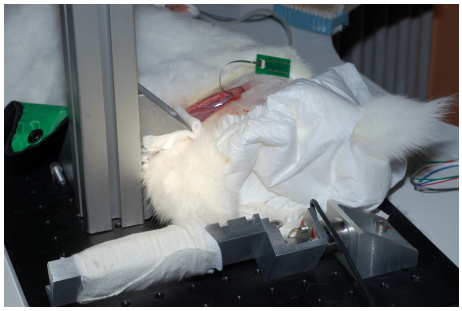

Fig. 7. Experimental setup for animal experiments.

To validate the complete toolchain, we plan to do an experiment on an animal model. We already have tested multipolar electrodes and stimulators using an experimental setup that we designed to measure the 3D strain and torque on a rabbit foot while stimulating the sciatic nerve. The knee of the animal is fixed, thus we consider only the ankle joint isometric force. This setup enables to try various strategies –standard, multipolar, anodal block, high-frequency block– and build recruitment curves for each one. We record EMG from the main muscles to have a direct view of their activation.

VI. Discussion and conclusion

We propose a toolchain to investigate, simulate and tune selective stimulation strategies. It relies mainly on OpenMEEG and Neuron with several contributions: the multipolar cuff electrode description and the computation of the potential at any position in the nerve in OpenMEEG, the axon model and its input in Neuron and finally the Python script which realises the interface between the two software.

The ongoing experiments will enable us to validate the models and the global toolchain. It will also help to tune the models to provide more accurate simulation. Then it will be possible to study and simulate selective stimulation strategies — either multipolar, anodal block or a combination of both — and test them experimentally.

There still are possible enhancements to these models. For the conduction model, anisotropic conductivities and non-cylindrical geometries would bring more realistic simulations, applicable to other types of cuff electrodes. We are working on the development of these extensions within OpenMEEG. Finally, the interface script is being extended to generate complex stimulation patterns in order to investigate the effect of the high frequency stimulation.

References

[1] A. V. Caparso, D. M. Durand, and J. M. Mansour, "A nerve cuff electrode for controlled reshaping of nerve geometry." *J Biomater Appl*, Nov 2008.

[2] K. Yoshida and K. Horch, "Selective stimulation of peripheral nerve fibers using dual intrafascicular electrodes." *IEEE Trans Biomed Eng*, vol. 40, no. 5, pp. 492–494, May 1993.

[3] A. Vuckovic and N. Rijkhoff, "Different pulse shapes for selective large fibre block in sacral nerve roots using a technique of anodal block: an experimental study." *Med Biol Eng Comput*, vol. 42, no. 6, pp. 817–24, 2004.

[4] X. Zhang, J. R. Roppolo, W. C. de Groat, and C. Tai, "Mechanism of nerve conduction block induced by high-frequency biphasic electrical currents." *IEEE Trans Biomed Eng*, vol. 53, no. 12 Pt 1, pp. 2445–2454, Dec 2006.

[5] D. Andreu, D. Guiraud, and G. Souquet, "A distributed architecture for activating the peripheral nervous system." *J Neural Eng*, vol. 6, no. 2, p. 26001, Apr 2009. [Online]. Available: http://dx.doi.org/10.1088/1741-2560/6/2/026001

[6] M. Clerc, R. Veltz, D. Guiraud, and J.-L. Divoux, "The 3d potential induced by functional electrical stimulation with multi-contact cuff electrodes : simulation and validation," in *IFESS 2008, 13th Annual Conference of the International Functional Electrical Stimulation Society*, 2008, pp. 331–333.

[7] J. Kybic, M. Clerc, T. Abboud, O. Faugeras, R. Keriven, and T. Papadopoulo, "A common formalism for the integral formulations of the forward EEG problem," *IEEE Transactions on Medical Imaging*, vol. 24, p. 12–28, jan 2005.

[8] S. Jacquir, J. Fruitet, D. Guiraud, and M. Clerc, "Computation of the electrical potential inside the nerve induced by an electrical stimulus," in *EMBC 2007: Proceedings of the 29th Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, 2007.

[9] M. L. Hines and N. T. Carnevale, "Neuron: a tool for neuroscientists." *Neuroscientist*, vol. 7, no. 2, pp. 123–135, Apr 2001.

[10] M. H. Brill, S. G. Waxman, J. W. Moore, and R. W. Joyner, "Conduction velocity and spike configuration in myelinated fibres: computed dependence on internode distance." *J Neurol Neurosurg Psychiatry*, vol. 40, no. 8, pp. 769–774, Aug 1977.