New Electrode Layout for Internal Selectivity of Nerves

Olivier Rossel, Fabien Soulier, Serge Bernard, Guy Cathébras

New Electrode Layout for Internal Selectivity of Nerves

Olivier Rossel, *student, IEEE member,* Fabien Soulier,
Serge Bernard, *IEEE member,* and Guy Cathébras *IEEE member*

LIRMM, Université Montpellier II - CNRS - INRIA, 161 rue Ada, 34392 Montpellier, France.
Email: Firstname.Lastname@lirmm.fr

***Abstract*—A nerve is an enclosed, cable-like bundle of peripheral axons. Each axon or set of axons carries neural afferent or efferent information. Many applications need to detect or record these specific nervous data inside the nerve but it is a big challenge. The main issue is to achieve a good selectivity inside the nerve without being invasive. In this context, we propose a new layout of multipolar electrode allowing a very high level of spatial selectivity. This electrode has a flat-interface electrode with an array of poles. The idea is to find the best value for the inter-pole distance and the most suitable post processing in order to both improve selectivity in the nerve and reject external parasitic signals. In this preliminary work, we put emphasis on the simulation of the action potential as a method to help the electrode specification.**

I. Introduction

The propagation of action potentials (AP) along the axons can be recorded via the electrical activity of the nerve (electroneurogramm, ENG). Unfortunately, this signal appears to be of very low level and even often below the micro-volt.

Moreover, bioelectrical activity makes the *in-vivo* environment very noisy, the worst noise being the signal generated by muscle activity (electromyogram, or EMG). In the particular case of peripheral nerve sensing, the EMG can exceed ENG by three order of magnitude at least. This parasitic signal will inevitably masks the ENG signal. Analog pre-processing must therefore be carried out in order to reject EMG-type noise.

The majority of ENG systems are dedicated to intracortical recording [1] where the electrode is an array of contacts inserted into the brain. The systems dedicated to ENG recording on peripheral nerves are often based on tripolar [2], [3], multipolar cuff [4] electrodes. The electrodes consist of three or more conductor rings around the nerve. Another kind of electrodes is based on a flat interface [5] where the central poles are placed perpendicularly to the propagation direction. These electrodes reach the best selectivity of the internal nerve activity. Nevertheless, it has been shown [5] that it is impossible to distinguish sources if they are closer than few millimeters. Based on this flat shape we propose a new multipolar electrode. To be more selective, this electrode is designed according to the nerve physiology and topology.

The first section of this paper gives an overview of the electrode specifications. The third section presents simulations of axons. Based on simulation results, the fourth and the fifth section present the proposed solution for preprocessing and electrode respectively. The last section gives some concluding remarks and perspectives of this work.

II. Overview of the Electrode Specifications

The propagation of neural signal along myelinated axons is due to the saltatory conduction. The action potential (AP) propagates along the axon from Nodes of Ranvier (NoR) to NoR (figure 1). When an AP occurs, several NoR are active at the same time, and the current density is localized around these active NoR. This local effect creates a voltage difference in the neighborhood of the NoR.

Solutions for neural activity recording are widely based on the dynamic features of these voltage differences due to the propagation of APs. In this paper, the pragmatical approach we propose consists in focusing firstly to the static sensibility of the considered electrode before studying the dynamic effects. In other words, before studying the dynamic neural activity (and the propagation), we propose to optimize the static sensibility of axon activities inside the nerve.

The main types of electrodes available are: cuff electrodes (flat or cylindrical), intrafascicular and sieves. Each kind has advantages and drawbacks, but for our application, we focus on the safety for the nerve and the selectivity of the electrode. By safety, at the electrode level, we mean the property of the electrode to be as non-invasive as possible for the nerve. Electrode selectivity is the ability for the electrode to isolate the activity of a particular set of axons inside the targeted nerve.

Because safety is the most critical issue, we choose to consider only cuff electrodes. This kind of electrode is not invasive for the nerve, and benefits of a large experience of long-term implantation on human beings [6]. Unfortunately, classical cylindrical cuff electrode has a very low selectivity because it gives global information averaged over the nerve surface. In the case of flat-interface electrode [5], the selectivity is enhanced but it is not possible to extract activity of single or small set of axons yet.

Our objective is to develop an electrode topology based on this flat shape electrode but with new characteristics. The resulting multipolar electrode is defined by the number, the type, the size and the layout of the poles around the nerve. The simulations presented in the next section will allow us to determine the best values of these electrode characteristics.

III. Simulation

The ENG signal can be described as the superimposition of extracellular potentials generated by several axons activated at the same moment. Our objective is to detect activity from

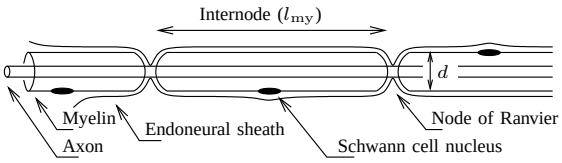

Fig. 1. Structure of a myelinated nerve fiber. Adapted from [9].

a set of these axons. We can assume that if the electrode characteristics are optimized to detect one axon activity then they can be extrapolated to be the best solution for a set of axons. We thus start by simulating an individual axon thanks to the *Neuron* simulation software [7].

A. Nerve Fiber Model

For the simulation, we consider the myelinated axon illustrated figure 1 where l_{my} is the distance between two successive NoR and d is the fiber diameter. According to the study by McIntyre *et al.* [8], l_{my} can vary from 0.5 to $1.5\,\mathrm{mm}$, with d varying from 5.7 to $16\,\mu\mathrm{m}$. We used these characteristics to build several *Neuron* models composed of 150 myelinated sections. The following simulation results focus on three different fibers, for a length of $l_{\mathrm{my}} = 0.5 - 1 - 1.5\,\mathrm{mm}$ with respectively a diameter of $d = 5.7 - 8.7 - 16\,\mu\mathrm{m}$.

B. Extracellular Potential Simulation

The action potential is triggered by synaptic current at one end of the fiber. To limit border artifacts, the electrode is placed near the middle of the nerve fiber model at a distance of about $2.5\,\mathrm{cm}$ from the starting point of the axone and $2\,\mathrm{cm}$ from the ending point of the axone. In order to estimate the contribution of local membrane currents to the total extra-cellular signal, we use the extension called *Extracellular Stimulation and Recording* available on the *Neuron* website[1]. The basic principle of this program is to computes the transfer resistances coupling the trans-membrane current to the recording site potential.

A typical result is shown on the figures 2 a1), a2), a3) for different fiber diameters 5.7,8.7 and $16\,\mu\mathrm{m}$, at a distance of $200\,\mu\mathrm{m}$ from the membrane. We can clearly distinguish the pseudo-periodical variations due to the discontinuities along the myelin shield. These variations are specific to the neuro-signal and can be highlighted by a spatial frequency analysis.

C. Spatial Frequency Analysis

The figures 2 b1), b2), b3) give spectrum for different fiber diameters 5.7,8.7 and $16\,\mu\mathrm{m}$, at a distance of $200\,\mu\mathrm{m}$ from the membrane. The frequencies of the highest amplitude value are clearly linked to the internode distance l_{my} and occur for wavenumbers equal to:

$$k_n = n/l_{\mathrm{my}}, \quad n \in \mathbb{N}, \tag{1}$$

whereas the low frequency energy denotes the global variations due to the signal propagation along the axon.

[1] `http://www.neuron.yale.edu`

It must be noticed that in *in vivo* environment, the main contribution of parasitic signal such as EMG would be in low spatial frequencies (at zero frequency for an uniform spatial influence). Our objective will be thus to reject this frequency domain in order to avoid any saturation of the electronic amplifier used for recording, while amplifying as much as possible the targeted neural signal. To achieve this goal a specific preprocessing on the signals recorded on the different poles is essential.

IV. Preprocessing

A. Principle

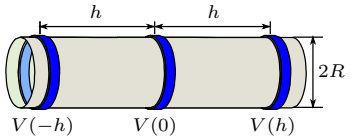

Fig. 3. Tripolar cuff electrode.

The classical preprocessing used in ENG recording is performed with tripole electrodes (see fig. 3). It consists in calculating the average of the potential differences between the central pole and each of the outer poles [10], [11]:

$$V_{\mathrm{filt}} = \frac{(V(0) - V(-h)) + (V(0) - V(h))}{2} \tag{2}$$

$$= V(0) - \frac{(V(-h) + V(h)}{2} \tag{3}$$

The last expression shows that this operation consists in applying a spatial high-pass filter calculating the second order derivative (Laplacian filter). Laplacian filters can reject both homogeneous and linearly varying potentials like those created by distant EMG sources [12].

This kind of processing can be extended to the case of multipolar electrode. For example, Rieger *et al.* [4] present an 11-pole electrode with the associated amplifier giving 9 laplacian-filtered outputs.

It is commonly admitted that the inter-pole distance should not be too small in order to not attenuate the signal of interest. But a genuine optimization of this distance may be achieved by studying the spatial filter frequency response.

B. Filter Characteristics

The frequency response of the filter can easily be expressed using the Fourier transform. We will assume that the poles are relatively small with regard to the wavelength of the signal (reflecting the variation of the potential). The potential is then ideally sampled by the pole distribution over the space and the frequency response of a laplacian filter composed of three poles located respectively at $x_n = -h, 0, h$ is:

$$H(k) = -\frac{1}{2}e^{2i\pi kh} + 1 - \frac{1}{2}e^{-2i\pi kh} \tag{4}$$

$$= 1 - \cos 2\pi kh. \tag{5}$$

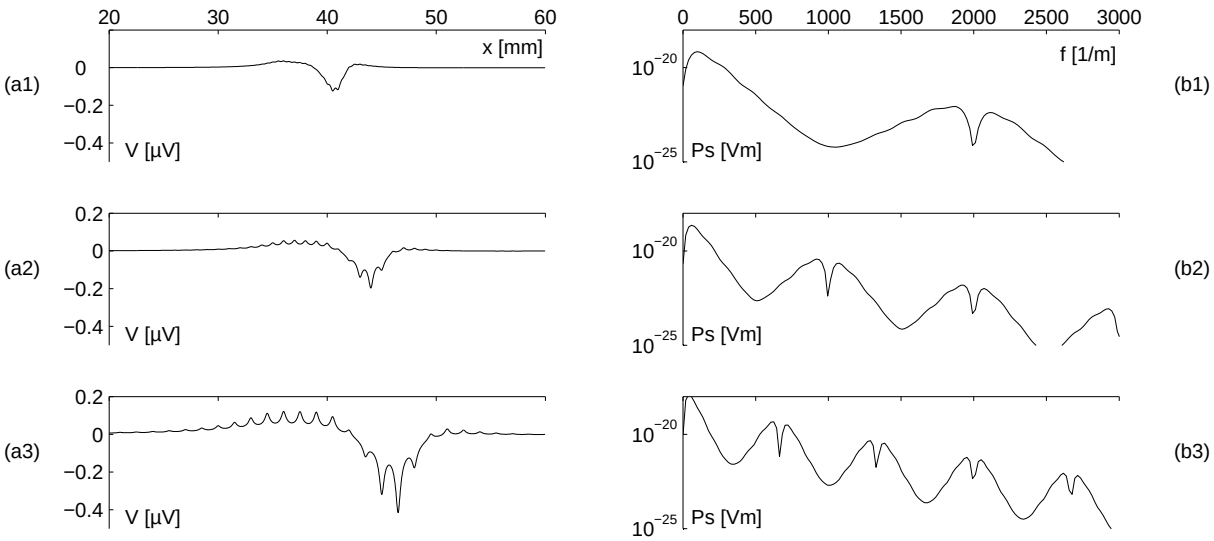

Fig. 2. Extracellular potential plotted along the axon's axis at a distance of $200\,\mu\mathrm{m}$ from the membrane, for different fiber diameters 5.7,8.7 and $16\,\mu\mathrm{m}$ respectively in **(a1)**,**(a2)** and **(a3)**.Typical spectrum of the extracellular potential regarding the spatial frequency in the axon's axis direction, for different fiber diameters 5.7,8.7 and $16\,\mu\mathrm{m}$ represented respectively in **(b1)**,**(b2)** and **(b3)**.

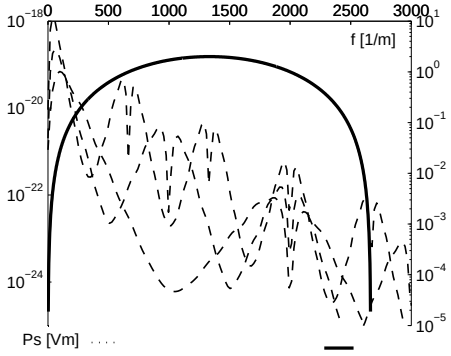

Fig. 4. Gain for the transfer function of our filter placed according to typical spectrum present in the fig. 2

This last expression shows that the Laplacian approximation (second order derivative) is only true for wavenumber $k << \frac{1}{h}$, when

$$H(k) \underset{0}{\approx} (2\pi^2 h)k^2 = -K(\mathrm{i}k)^2, \tag{6}$$

since $\mathrm{i}k$ denotes the spatial derivative. In this equation, low frequencies (potential created by distant sources) are obviously rejected with a $40\,\mathrm{dB/dec}$ cutoff. But it can easily be seen on the real frequency response (equation 4) that the filter presents a $6\,\mathrm{dB}$-bandwidth between $1/4h$ and $3/4h$ (fig. 4).

The first bin of the ENG spectrum is centered on $k = 1/l_{\mathrm{my}}$ (fig. 4) for $0.5\,\mathrm{mm} < l_{\mathrm{my}} < 1.5\,\mathrm{mm}$. To fit the bandpass of the filter for all kind of fiber, the optimal inter-pole distance should simply be

$$h = 375\,\mu\mathrm{m}, \tag{7}$$

which is a surprisingly small interval. The result of this filtering is shown on the fig. 5.

V. Electrode Design

A. Introduction

As previously explained, the preprocessing used in classical ENG systems to reject parasitic signal like EMG is based on a Laplacian filter. This is the same preprocessing as we propose but in these cases each pole is a complete ring or at least the size of each pole is very large in comparison to the distance between two NoR. Moreover, the width of poles acts like a low-pass filter (in spatial domain) by averaging the potential. In this context, they offer very poor spatial selectivity and are not really suitable for source separation. Therefore, we propose an entirely new design of multipolar cuff electrode using very small pole distributed all over the cuff. The figure 6 gives the top and the cross section view of this kind of electrode. Previous works have shown that this kind of multipolar design persents both better sensitivity and greater selectivity than the tripolar cuff with regard to the potential created by a single NoR [12]. We will consider only six poles of this electrode as illustrated fig. 6. The objective is to evaluate the sensitivity of this set of poles versus type and depth of the axon.

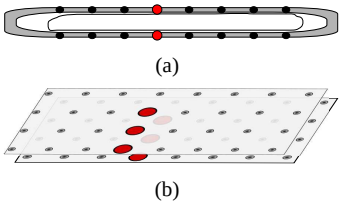

Fig. 6. Multipolar electrode cuff model.

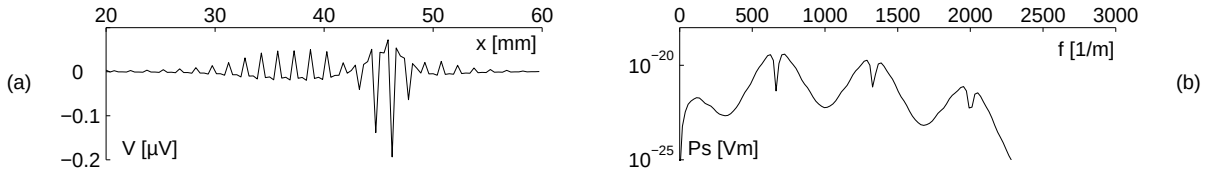

Fig. 5. Laplacian potential plotted along the axon's axis at a distance of $200\,\mu\mathrm{m}$ from the membrane, for fiber diameter $16\,\mu\mathrm{m}$ **(a)**. Typical spectrum of the Laplacian filter output versus the spatial frequency into the axon's axis for this fiber **(b)**.

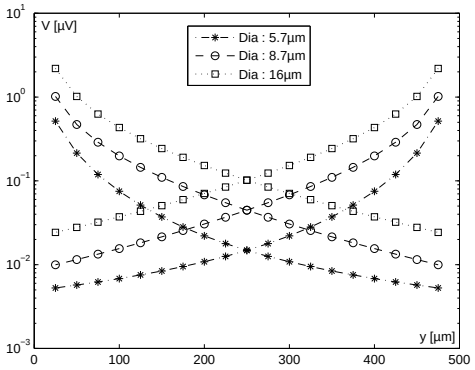

Fig. 7. Maximal measurable voltages according to the depth of the axon from the 6 pole electrode for various axon diameters.

B. Sensibility versus Type and Depth of the Axon

To evaluate the sensibility we perform two Laplacian filtering on 3 poles : one on the top and another one on the bottom of the electrode (see fig. 6). The figure 7 gives the maximal values in the two filter outputs for different axon diameters and for a flat-interface electrode with a thickness of $1/2\,\mathrm{mm}$.

Obviously, larger and/or closer axons are easier to detect. We can notice that the attenuation of the measured signal due to the distance between 3 poles (top or bottom) and the considered source (the axon) becomes quickly significant. But when this distance is higher than 0.25mm the opposite 3 pole motif sensitivity becomes higher and could be used to detect axon activity. Moreover, this high variation of the signal attenuation means that each 3 pole motif would be only sensitive to fibers close to it. In other words this type of electrode has a high spatial selectivity (about $1/2\,\mathrm{mm}$).

VI. Conclusion and Perspectives

This premilinary work has presented an ENG recording electrode design. The two main principles are the use of a flat-interface electrode with numerous poles and the Laplacien filtering on longitudinal poles. The simulations have shown that the optimal distance between the poles for this type of electrode is about $375\,\mu\mathrm{m}$. This inter-pole distance is much less than the classical distance between poles in multipolar cuff electrodes. These first results allow us to explore another kind of multipolar electrodes based on distributed poles with small inter-pole distance. This new type of electrodes allows us to have a new view on the ENG signal. We do not consider only the propagation of the action potential, but also the Nodes of Ranvier (NoR) locations.

An experiment is in progress, to prove that electrodes with short inter-pole distance can measure the influence of NoR locations and to evaluate signal to noise ratio. This experiment will be set up on a worm because of its really simple neural system consisting of only three axons, and having pseudo NoR. For a complete study on selectivity the next work will consist in creating a realistic model of a complete nerve based on classical distributions and properties of axons.

References

[1] T. Jochum, T. Denison, and P. Wolf, "Integrated circuit amplifiers for multi-electrode intracortical recording." *Journal of Neural Engineering*, vol. 6, no. 1, p. 012001 (26pp), January 2009.

[2] K. Papathanasiou and T. Ehmann, "An implantable CMOS signal conditioning system for recording nerve signals with cuff electrodes," in *ISCAS'2000 International Symposium on Circuits and Systems.*, vol. 5, Geneva, 2000, pp. 281–284.

[3] J. Nielsen and E. Bruun, "A low-power CMOS front-end for cuff-recorded nerve signals," in *Proceedings of the 22nd Norchip Conference.*, Nov. 2004, pp. 24–27.

[4] R. Rieger, M. Schuettler, D. Pal, C. Clarke, P. Langlois, J. Taylor, and N. Donaldson, "Very low-noise ENG amplifier system using cmos technology," *IEEE Transactions on Neural Systems and Rehabilitation Engineering*, vol. 14, no. 4, pp. 427–437, December 2006.

[5] P. B. Yoo and D. M. Durand, "Selective recording of the canine hypoglossal nerve using a multicontact flat interface nerve electrode," vol. 52, no. 8, pp. 1461–1469, 2005.

[6] R. L. Waters, D. R. McNeal, W. Faloon, and B. Clifford, "Functional electrical stimulation of the peroneal nerve for hemiplegia: long-tem clinical follow-up," *Journal of bone and joint surgery*, vol. 67, no. 5, pp. 792–793, 1985.

[7] N. Carnevale and M. Hines, *The Neuron Book*. Cambridge University Press, 2006.

[8] C. C. McIntyre, A. G. Richardson, and W. M. Grill, "Modeling the excitability of mammalian nerve fibers: Influence of afterpotentials on the recovery cycle," *Journal of Neurophysiology*, vol. 87, pp. 995–1006, february 2002.

[9] R. Plonsey and R. C. Barr, *Bioelectricity: A Quantitative Approach*, 3rd ed. Springer, June 2007.

[10] J. J. Struijk and M. Thomsen, "Tripolar nerve cuff recording: stimulus artifact, EMG and the recorded nerve signal," in *Engineering in Medicine and Biology Society, 1995. IEEE 17th Annual Conference*, vol. 2, Montreal, Que., September 1995, pp. 1105–1106.

[11] C. Pflaum, R. R. Riso, and G. Wiesspeiner, "Performance of alternative amplifier configurations for tripolar nerve cuff recorded ENG," in *Engineering in Medicine and Biology Society, 1996. Bridging Disciplines for Biomedicine. Proceedings of the 18th Annual International Conference of the IEEE*, vol. 1, Amsterdam, 1996, pp. 375–376.

[12] F. Soulier, L. Gouyet, G. Cathébras, S. Bernard, D. Guiraud, and Y. Bertrand, "Multipolar electrode and preamplifier design for eng-signal acquisition," in *Biomedical Engineering Systems and Technologies*, ser. Communications in Computer and Information Science, A. Fred, J. Filipe, and H. Gamboa, Eds. Berlin, Heidelberg: Springer, 2009, vol. 25, ch. Multipolar Electrode and Preamplifier Design for ENG-Signal Acquisition, pp. 148–159.