

A Steering Electrode Array for Selective Stimulation of Sacral Nerve Roots

Fabio O. Rodrigues, Paulo M. Mendes, Marian Bartek, Benjamin Mimoun

Abstract— In this work a cylindrical electrode array to be used for electrical stimulation of sacral nerve roots is studied in respect to its ability to achieve selective stimulation of various spatial regions of the nerve bundle. Simulation results achieved on a simplified model consisting of 6 electrodes evenly distributed along the sacral root perimeter indicate that this goal is feasible.

A demonstrator of the proposed electrode array was designed and is implemented on an ultra-thin polyimide foil that is to be rolled around the sacral root bundle and then fixed adhesively. Multichannel steering electronics connected through the rigid silicon chip will be used to perform in vivo animal tests for the proof of concept.

Index Terms—Functional electrical stimulation, selective neural stimulation, steerable electrodes, spinal cord injury.

I. INTRODUCTION

Spinal cord injuries (SCI) affect millions of people. Disorders affecting normal urinary functions like micturition and continence are experienced as the most devastating for the daily life of these patients. As restoration of control and function of the lower urinary tract in SCI-patients to the pre-injury state is not yet possible, the goal of treatment is to prevent the urological complications while preserving continence. Effectively, catheters are still the most used technique concerning bladder management [1]. However, the high incidence of urinary tract infections associated with catheter use presents a continuous burden on patients and the medical system. Furthermore, many patients are not able to catheterize themselves or are not willing to depend on catheterization, as it can be physically or socially uncomfortable.

II. FUNCTIONAL ELECTRICAL STIMULATION FOR BLADDER CONTROL

In terms of physiology, bladder emptying would be closest to normal when induced by a detrusor (bladder wall muscle) contraction. Electrical stimulation can be a good solution to achieve such contraction. In fact, during the last four decades, several different places have been studied in order to evoke bladder contraction, via electrical stimulation of specific

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nerves or also directly on the detrusor muscle itself [2]. Fig. 1 shows the main nerves and muscles involved in the urinary bladder voiding. Regarding all the different stimulation sites available for bladder stimulation, sacral nerve roots remain the optimal site for clinical application [3]. The space within the spinal column (subarachnoid space) facilitates mechanically-stable electrode positioning and is also an easy place to access during the surgery.

Sacral nerve roots have a mixed population of nerves: somatic fibers innervating the urethral sphincter and parasympathetic fibers innervating the detrusor muscle. One main difference between these two kinds of fibers is that the somatic are larger (in diameter) than the parasympathetic ones and once threshold is lower for larger fibers, it means a smaller stimulus is needed to activate the parasympathetic fibers and consequently to activate the sphincter. So, if an electric current is uniformly applied to the sacral nerve root, bladder and sphincter will contract simultaneously and no voiding occurs. *Brindley et al.* proposed a method to overcome the simultaneous activation of both detrusor and sphincter [3]. Basically, it consists in the different relaxation times of these muscles. So, when the sacral nerve root is stimulated, both muscles contracts but sphincter (because it is a striated muscle) relaxes faster than the detrusor, allowing voiding. However, this micturition process occurs is spurts and it is not “natural.”

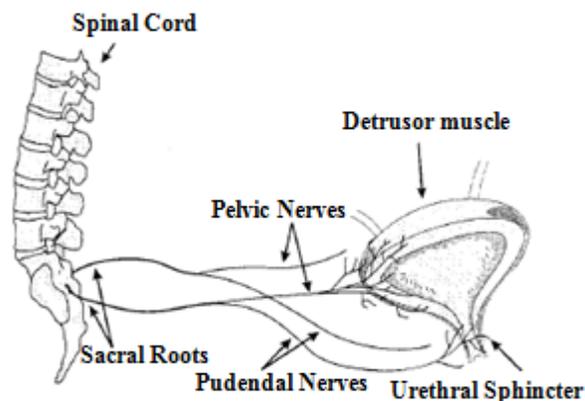


Fig. 1 Nerves and muscles involved in the bladder voiding process.

As mentioned above, sacral nerve roots are a mixed bundle of different fibers, normally called as rootlets. These rootlets have different nature, as they come from different neuronal structures inside the spinal cord and they also innervate different muscles in the lower part of the body – lower limbs,

rectum, bladder, sphincter, penis, etc. Normally, in humans there are 5 sacral nerve roots (S1-S5), and each one of them can have several rootlets. As these rootlets are highly specialized – preferably one rootlet is somatic or it is parasympathetic – if one of them is selectively stimulated, its effect will be dominant [4]. This principle can be used to selectively recruit the parasympathetic fibers innervating detrusor muscle inside the sacral nerve roots.

The goal of this study is to evaluate if it is possible to steer the electrical field in such a way that selectivity can be accomplished in the sacral nerve root stimulation. In order to achieve this, a multipolar cuff electrode array is proposed. Similar concept and electrode arrays have been used for deep brain stimulation and have been shown to be successful in achieving selectivity [5].

III. METHODS

The activation of a nerve fiber by external stimuli has been researched since the 1930's. Important models were reported on how the membrane of the nerve fiber can be modulated by an electrical circuit and which kind of stimulus can be applied to generate an action potential in a nerve fiber [6][7]. Effectively, the nerve fiber starts an action potential when the second partial derivative of the electrical potential in the direction of the neural fiber length is higher than a certain threshold value ($\partial^2 V / \partial z^2 > k$) [6].

In the present work, the evaluation of the potential for the proposed electrode array was performed using *Ansys Multiphysics v12.0*. Part of the model is presented in Fig. 2.

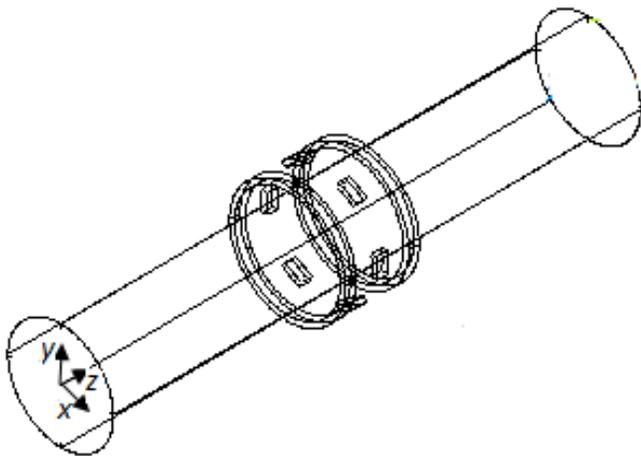


Fig. 2 Schematic model of the nerve fiber together with the electrode array used in simulations. Note the coordinate system x , y , z .

The electrode array consists of 6 electrodes at the inner part that can be used as cathodes (or also as anodes), and 2 anode ring electrodes at the outer part. All the electrodes are $50 \mu\text{m}$ thick and $100 \mu\text{m}$ wide. The separation between the outer and inner electrodes is $200 \mu\text{m}$. The length of the 6 inner electrodes is $150 \mu\text{m}$. In this model, the sacral nerve diameter used is 1 mm (the typical diameter for the S2 sacral roots) and the modeled section is 5 mm long.

Besides nerve and the electrodes, the model includes also an insulating carrier film (to avoid current leakages) and a

cerebrospinal fluid (CSF). These are not shown in Fig. 2. At the model boundary the Dirichlet boundary condition ($V=0$) was applied. In Table 1 all material properties (electrical conductivities) are summarized.

Table 1. Electrical conductivities of the media used in the model. Note that only the nerve bundle has different conductivities in the xy plane and along the z -axis.

Model compartments	Conductivity [S/m]	
	σ_{xy}	σ_z
Nerve	0.083	0.6
Insulating film	0.0017	0.0017
Electrodes	6	6
CSF	1.7	1.7
Boundary	0.01	0.01

All the compartments are isotropic except the nerve with conductivity σ_z in a direction parallel to the bundle axis and σ_{xy} perpendicular to this axis. The conductivities of nerve and CSF are taken from Geddes and Baker [8]. The other two parts, i.e., the insulating film and the electrodes, have been given a low and a high value of electrical conductivity, respectively.

The Finite Element (FE) model was meshed using the 20 node electric solid element *SOLID231*. The resulting mesh shows still some inaccuracies and need to further be optimized. But it is sufficient for the initial results.

IV. RESULTS

As it was already mentioned, the activation of a nerve fiber depends on the second partial derivative of the potential along the length of the fiber (z -axis in this model). In this work, only the simplest case where one of the inner metal electrodes is used as a source (electric current injection) and all other electrodes are grounded is studied. In Fig. 3, all the electrodes are shown and the inner electrodes are labeled with letters (A-F). The *Electrode A* was used to inject current into the nerve volume.

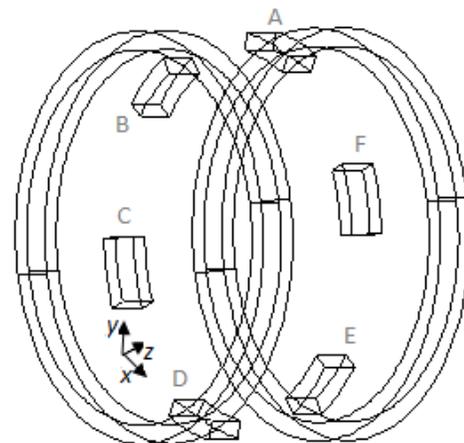


Fig. 3 Electrode array geometry: 2 ring anodes and 6 inner electrodes evenly distributed on a circle perimeter are available for charge injection. In this study *Electrode A* was used as a cathode.

In order to evaluate the selectivity of this electrode, 2 different simulations were performed. The only factor changing between these two simulations was the current injection in A: 0.9 mA and 1 mA. Then, a plane containing z axis and passing through *Electrodes A* and *D* was set. On this plane *Line 1* was selected in order to present the 2nd partial derivative in z-direction for the two current values. *Line 1* is set at 80 % of the cylinder radius measured from its centre. The simulation results are shown in Fig. 4.

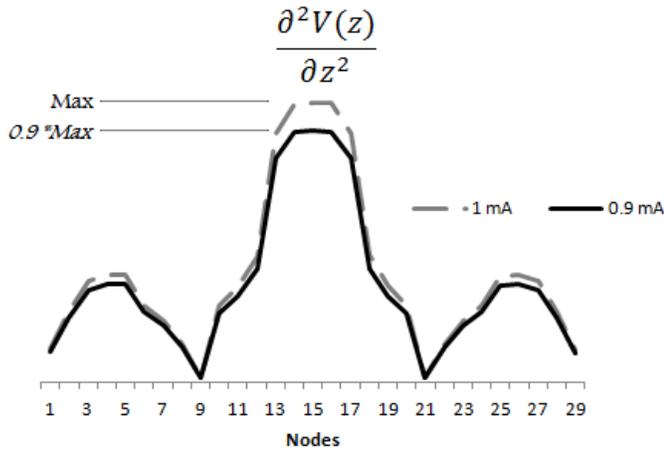


Fig. 4 Second derivative of the potential along *Line 1*. (*Line 1* is parallel to z-axis; at 80 % of the cylinder radius). Node 1 corresponds to the first node at the left anode. Node 29 corresponds to the last node at the right anode. Nodes 13-17 correspond to the z coordinates of the inner contacts.

From Fig. 4 it can be concluded that the second partial derivative of the electrical potential along z axis has its maximum in the region under *Electrode A*. However, the most important issue is how the activation region varies with varying the injection current. Only if this region can be well defined, this technique is suitable for selective control. For this reason, the difference in maximum values of activation potential for different current values was studied. As can be seen in Fig. 4, by varying the stimulation current between 0.9 and 1 mA selective stimulation could be achieved. If we assume a fiber at the location of *Line 1*, with a threshold value (to start an action potential) of about 0.95 times the maximum, we know this fiber will only be activated with current of 1 mA but not with 0.9 mA.

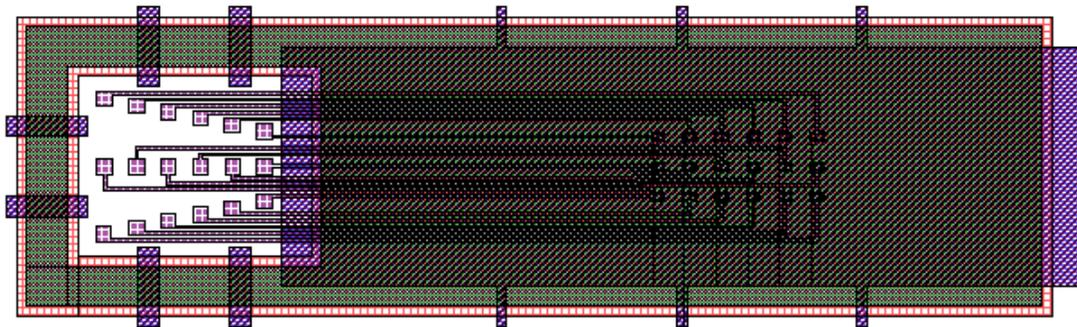


Fig. 6 The flexible steering electrode array is implemented in a “flex-to-rigid” technology using a 5 mask process. The silicon chip is in white with the bondpads on its surface. The interconnects and the aluminum electrode contacts are also shown.

In Fig. 5 the isopotential lines in the plane containing the z-axis and crossing *Electrodes A* and *D* for a 1 mA current injected into *Electrode A* are shown.

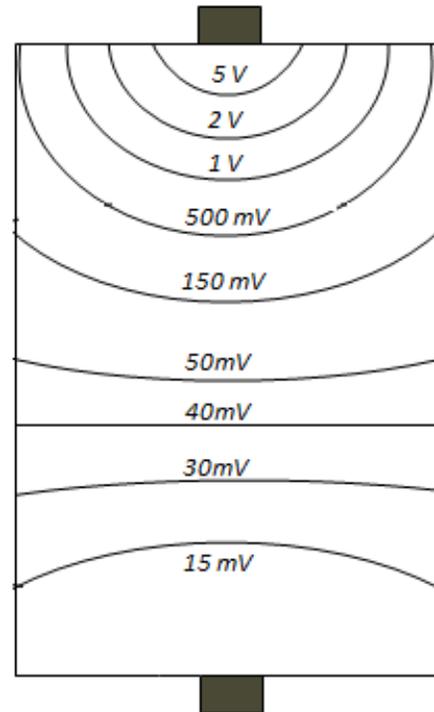


Fig. 5 Isopotential lines on the plane containing z-axis and crossing *Electrodes A* and *D*. An electric current of 1 mA is injected into *Electrode A* (black rectangular at the top); *Electrode D* at the bottom serves as an anode.

V. PROTOTYPE FABRICATION

The prototype electrode array with 18 electrodes (similar to the one presented here) was designed and is being fabricated using “flex-to-rigid” technology [9]. The design masks are shown in Fig. 6. The activation electrode array is implemented on an ultra-thin polyimide foil that is to be rolled around the sacral root bundle and then fixed adhesively.

VI. CONCLUSION AND FUTURE WORK

A cylindrical electrode array for selective stimulation of sacral nerve roots was presented. The results obtained from a finite element analysis using a simplified electrode array model indicate that spatial selectivity is feasible. However further analysis and experimental proof of the concept is required.

In the future work, this basic study will be extended to more general distribution and higher number of the stimulation electrodes. Also, the influence of neural fiber diameter on the activation region will be included.

A prototype electrode array with 18 electrodes to be implemented on a flexible polyimide foil carrier was designed and is being fabricated. Multichannel steering electronics connected through the rigid silicon chip will be used to perform *in vivo* animal tests for the proof of concept.

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