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# „A Virtual Workbench For Peripheral Electrode Design“

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## Introduction

Cuff-electrodes are the most widespread peripheral nerve interface for recording and stimulation in animal models. Especially in the early phase of experimental series the specifications for the electrodes are subject to change, due to e.g.: varying nerve diameters or insufficient noise reduction. These adaptations include geometry (length, diameter, arrangement of electrodes, etc.) as well as physical parameters (transfer function, maximum charge delivery as a function of electrode material, etc.). The micromachining of thin film electrodes, however, is not suitable for rapid changes in the design due to lithography masks and corresponding clean room processes for both, time and process costs.
In order to allow faster prototyping and testing of specific functionality of electrodes a virtual testing environment is needed to simulate properties of an electrode design on a peripheral nerve prior to manufacturing. Target specifications of such a virtual testing environment include:

- robust algorithms for simulation of transfer function
- simple changes of geometrical electrode parameters
- inclusion of realistic electrical and structural biological model parameters
- realistic simulation of specific and local stimulation
- realistic simulation of neural excitation / activity

Especially those topics that resemble biological parameters are quite demanding and can only be implemented in a simplified manner. Using different software packages we were able to realize a virtual workbench that fulfills all of the above-mentioned topics.

## Materials and Methods

### *Modeling of simplified neural structure for stimulation*

The software package COMSOL (Comsol Multiphysics) allows the simulation of electrical fields and currents in defined materials. One of our nerve models (Figure 1) consists of a cylinder (1 mm diameter) and contains 20 simulated nerve fascicles. For the model, the typical dielectric constants and conductivities discussed in [1] were used. The electrodes were implemented as highly conductive rings and blocks, which carry the stimulation currents. The number of fascicles per neuron (and thus the size of the nerve) and the arrangement of fascicles inside the nerve can be set (Fig. 2) and the stimulation arrangement can be included, e.g. steering stimulation [2].

### *Modeling of simplified neural structure for recording*

Cuff-electrode sites are too far off from the source of a firing axon to record single- or multiunit activity. However, we use a compound action potential (CAP)-based approach to simulate the net sum activity of our simplified nerve model. In the literature models using COMSOL for simulation of a single neuron activity basically rely on the Fitz-Hugh-Nagumo model (FHN) as an adaption for the spreading action potentials of the Hodkin Huxley (HH) cycle [3].

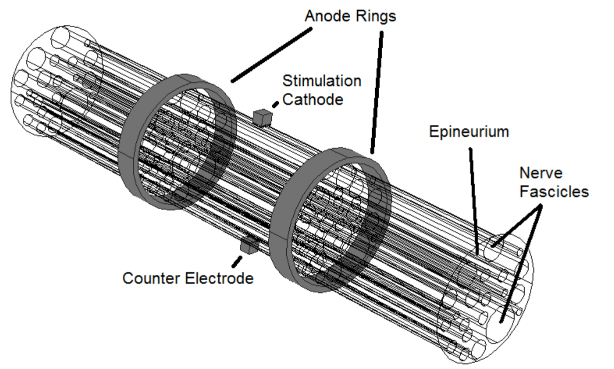


**Figure 1 Simplified nerve model with cuff-electrode implementation for stimulation**

Since our model involves many fibers, our approach uses a simplified 3D model and contains the following features:

- Fourier series adaptation (instead of FHN) of the HH cycle to resemble the AP's of individual fibers
- AP propagation follows salutatory conduction in myelinated fibers
- spacing between nodes of Ranvier can be adjusted between 0.2 sand 2.0 mm
- size of nodes can be changed according literature values
- fibers of ≤ 1 µm are considered C-fibers with electrotonic conduction
- AP proliferation speed depends on diameter of the fiber following values of the Erlanger-Gasser model
- implementation of electrical attenuation parameters of neural tissue (Fig. 3)
- Poisson distributed activity pattern of "background" neurons

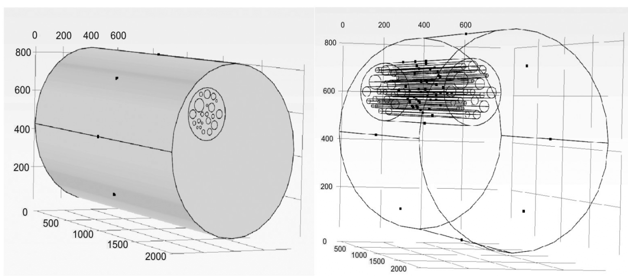

**Figure 2 Alternative setting of simulated nerve.**

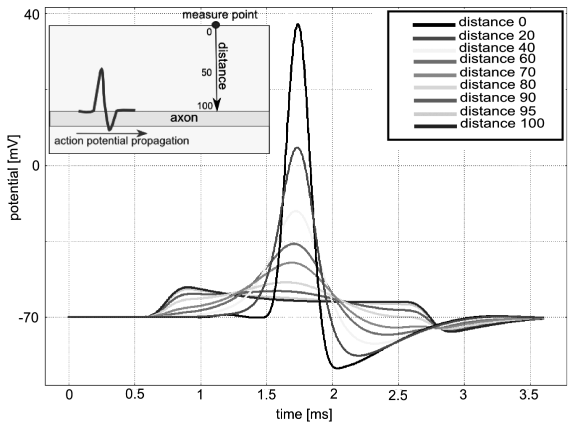


**Figure 3 Attenuation of electrical field in vertical plane.**

## Results

Different recording and stimulation paradigms were realized in the simulation environment. Exemplary, we present specific and local stimulation results using steering currents. Chemical safe stimulation limits (charge injection capacity) of the driving electrode and its depth effect underneath were taken into account. If important fibers are localized in deeper layers, steering currents can be applied to increase the depth of the super threshold current underneath the stimulation cathode. However, steering currents have to be chosen with care in order to avoid additional and unwanted stimulation via "virtual electrodes". Simulations include stimulation patterns, the size and the position of the electrodes.

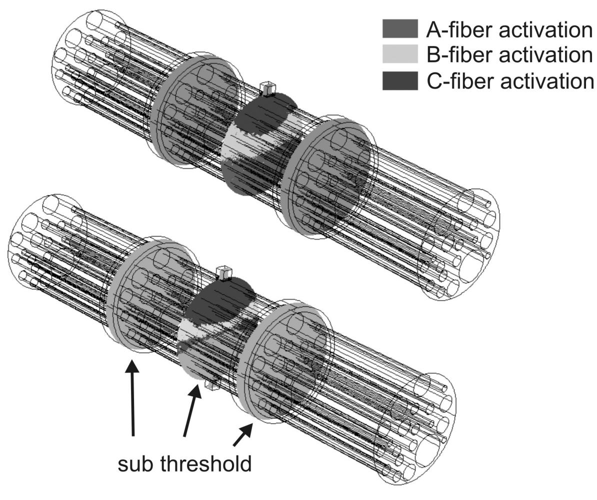


**Figure 4 Simulated distribution of fiber activation for simple and steering configuration at identical stimulation currents.**

In our example we compared the effect of a simple stimulation vs. a stimulation using steering currents (Fig. 4). Using identical stimulation in the upper electrode, the simple stimulation leads to a local activation of C-fibers, but A-fibers are in addition activated throughout the full diameter of the nerve, whereas the additional steering from a counter electrode leads to a more homogenous distribution of the stimulation thresholds across the diameter of the nerve. The steering stimulation is therefore more precise and local for the inversely recruited classes of fibers. If the diameter and roughly the composition of a nerve is known a virtual electrode can be arranged such that depending on the chosen stimulation protocol each electrode stimulates a defined sector of the nerve diameter while the steering controls the penetration depth.

## Discussion

The development of a realistic simulation model including the technical side of neural interfaces is a huge and challenging task. Our model is far from resembling the actual situation in a real nerve. Nevertheless it allows testing of design parameters for electrodes regarding specific functionalities prior manufacturing in a 3D model. Stimulation thresholds can be optimizes by adjustment of stimulation pattern or electrode configuration in the model. 3D reconstruction of peripheral nerves using histology methods is under work. This more realistic nerve model will become the future base for our simulation environment. However, this “real nerve” model will need more powerful computation power from grid based computation approaches.

## Conclusion

Our simulation environment allows a swift overview on the recording- and stimulation properties of new and established electrode design varying various parameters with respect to spatial and fiber selectivity for peripheral nerve interfaces.

## Literature


[1] S. Raspopovic, M. Capogrosso,and S. Micera (2011). A computational model for the stimulation of rat sciatic nerve using a transverse intrafascicular multichannel electrode *Ieee Transactions On Neural Systems And Rehabilitation Engineering*, *19*(4), 333–344.

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[3] R. Appali, S. Petersen, J. Gimsa, and U. van Rienen (2009). 3D-Simulation of Action Potential Propagation in a Squid Giant



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