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AN ARTIFICIAL NERVE FIBER FOR EVALUATION OF NERVE CUFF ELECTRODES

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Abstract - **The different applications of natural sensors for feedback in rehabilitation systems using functional electrical stimulation (FES) require specialised and optimised designs of nerve cuff electrodes for recording of the sensory information. This paper describes a simple artificial nerve fiber for evaluation of nerve cuff electrode designs, cuff recording configurations and noise reduction methods in a controlled environment. The idea is to experimentally identify the transfer function from a nerve fiber to the cuff electrode and make a computed construction of a single fiber action potential. This is done using an action current template as input to the identified transfer functions. The artificial nerve was tested on three cuffs of different lengths and the results showed good agreement with the results from a nerve conduction model of a single fiber action potential in an inhomogeneous volume conductor.**

INTRODUCTION

During the last years, there has been an increasing interest in the use of natural sensors for feedback in functional electrical stimulation (FES) assisted rehabilitation systems. The natural sensors are applicable in several areas of rehabilitation, such as restoration of hand grasp, drop foot correction, and control of standing [1,2,3]. The nerve cuff electrode has proved to be useful for recording of the electroneurogram (ENG) from sensory nerves. Each application area within FES imposes its own criteria for nerve cuff electrode design and specialised cuffs have to be developed.

Among the parameters that affect the ENG signal recorded with a cuff electrode are cuff dimensions, cuff closing methods, recording configurations, and methods to reduce muscle interference and stimulus artefacts. Evaluation of these parameters in acute animal experiments is a time consuming task and the results are often difficult to reproduce. Therefore, we have designed a simple artificial nerve fiber for evaluation of nerve cuff electrodes. The advantages of using the artificial nerve fiber instead of animal experiments are: Animals are saved, faster evaluation of cuff performance, more reproducible results and lower costs.

In the following, the use of the artificial nerve is demonstrated in the simulation of single fiber action potentials (SFAPs) for monopolar cuff electrodes of different lengths.

THEORY AND METHODS

For our purposes, a single myelinated nerve fiber can be represented as point current sources, distributed at the nodes of Ranvier [4] (Figure 1).

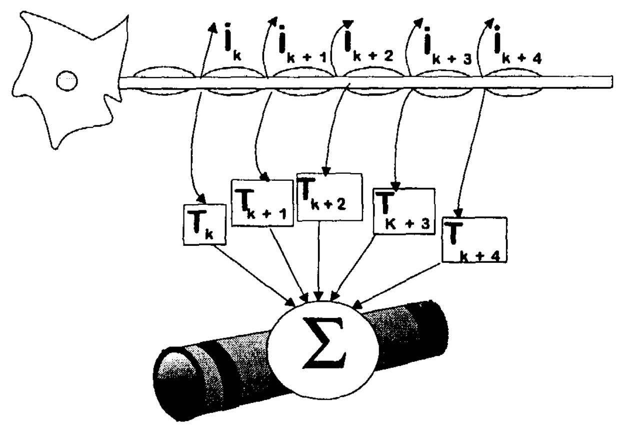

Figure 1 The recorded SFAP is the sum of the effects of each of the current sources that represent the nodes of Ranvier.

When an action potential is propagating along the nerve fiber, each node of Ranvier will generate a current with the shape of the tripolar action current wave, i_{AC}, (Figure 2), but the wave will be delayed by the time it takes the action potential to propagate to the respective node. From node to node the delay t_d is given by

$$t_d = \frac{L}{v_c} \qquad (1.)$$

where L is the internodal distance and v_c is the conduction velocity, which was assumed to be linear with fiber diameter, d: $v_c=5.58\text{x}10^6 d$ [4]. The action current template was calculated using a cable model, where the membrane kinetics of the nodes of ranvier were based on a modification [4] of the work of Chiu et al. [5]. The amplitude of i_{AC} is linear with fiber diameter.

Because an action potential is spread out over several nodes of Ranvier, the signal that is recorded with a cuff electrode is the superposition of each of the active point current sources (nodes of Ranvier) multiplied by their respective transform functions, T_k, from node k to the cuff electrode (see fig. 1). T_k is defined as

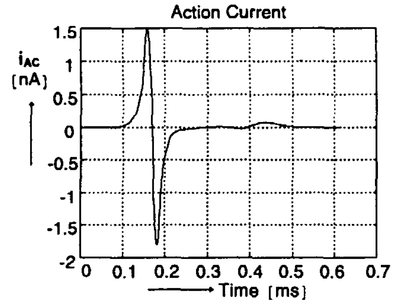

***Figure 2** The action current template, for a 10 µm diameter fiber, used for calculation of the SFAPs* [4].

$$T_k = \frac{V_k}{i_k} \tag{2.}$$

where V_k is the voltage that would be recorded if only node k is active with a current i_k. With (1.) and (2.) the recorded signal can be described as:

$$SFAP(t) = \sum_{k=1}^{nr.\ of\ nodes} i_{AC}(t - kt_d) \cdot T_k \tag{3.}$$

The transfer functions, T_k, for different cuffs were measured directly in an experimental setup shown in figure 3. The nerve cuff electrode was placed in a container with saline (22 °C). A wire of teflon coated stainless steel (diameter=300 µm), carrying the point current source, was moved through the cuff electrode with a stepper motor with steps of 0.5 mm. For each position, that is for each step of the stepper motor, a sine wave current, i_{in}, was applied to the artificial nerve and the resulting cuff potential, V_{out}, was recorded. The second current electrode was in the saline outside the cuff.

We assumed that the transfer functions are independent of frequency within the bandwidth of the neural signal, and therefore, the transfer functions were identified only for a fixed frequency: A current of 4.5 µA amplitude sine wave at 900 Hz was chosen, since our own recordings of ENG in humans have shown to peak around 900 Hz (unpublished).

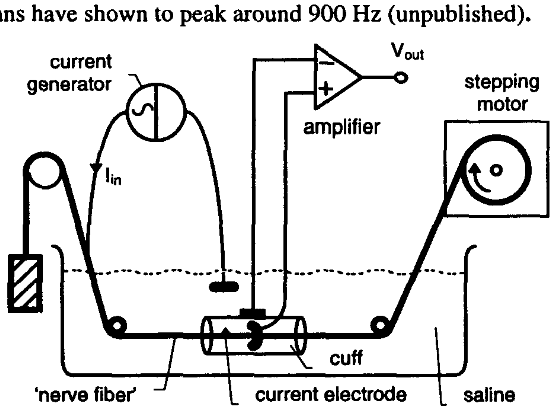

***Figure 3** The set-up for the artificial nerve fiber*

The recorded cuff signal was amplified 320 times and the applied current and the cuff signal was sampled with 25 kHz. Before computer construction of the SFAP, both the applied sine current and the recorded cuff signal were band pass filtered (850 Hz - 950 Hz) with a 3rd order Chebyshev filter. Then the transfer functions were calculated as the ratio of the RMS value of the potential and current, and used in equation 3 together with the action current template shown in Figure 2.

RESULTS

Measurements were done on three monopolar silicone nerve cuffs (inner diameter 2 mm, cuff lengths 5, 10, and 29 mm, respectively) with platinum electrodes in the middle of the cuff. Figure 4 shows the transfer functions T as a function of the position of the "node of Ranvier", and the constructed monopolar cuff recordings of a single fiber action potential (10 µm fiber diameter) for each of the three cuffs. For comparison, the calculated action potentials from a nerve conduction model of an inhomogeneous volume conductor [4] using the same cuff dimensions are also shown. In the transfer function plots the results of three trials are shown.

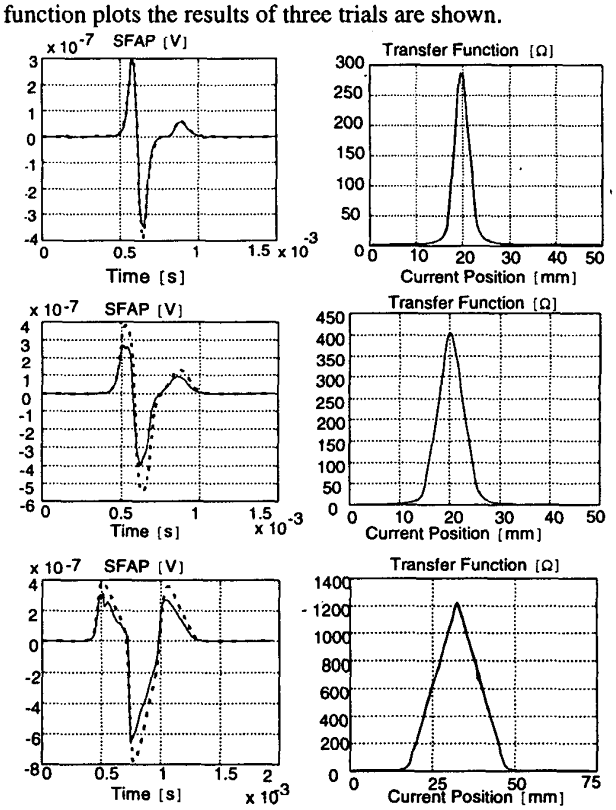

***Figure 4** Left column: solid: the SFAP as obtained with the artificial nerve; dotted: the results from a nerve conduction model using the same cuff and fiber dimensions* [4]. ***Right column:** the measured transfer functions plotted for three trials, for each cuff. **Top row:** Cuff length = 5 mm, **Middle row:** Cuff length =10 mm, **Bottom row:** Cuff length =29 mm.*

DISCUSSION

The SFAP obtained with the artificial nerve fiber is comparable with the predicted SFAP from the nerve conducting model, and there is a good reproducibility of the identified transfer functions and the SFAPs. This was the case for all three cuffs.

The smaller amplitudes of the artificial nerve signals compared with the modelled signal could be due to difference in the conductivity of the saline (at 22 °C) and the conductivity used in the model, which was specified for 20 °C [6].

The method is flexible in the sense that SFAPs from different diameter nerve fibers can easily be determined with a single measurement of the transfer function, by interpolation of the transfer function to the positions of the "nodes of Ranvier" of other diameter fibers. The stepping distance of 0.5 mm was sufficiently small to be able to do these interpolations reliably.

Moreover, the method is applicable to cuff configuration evaluations, tests of noise reduction methods, and compound action potentials. The artificial nerve fiber is thus a useful tool in the study of the effect of different cuff design parameters on the recorded signal. We recommend this simple approach as a low cost alternative to acute animal experiments.

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