MULTI-ELECTRODE NERVE CUFF RECORDING – MODEL ANALYSIS OF THE EFFECTS OF FINITE CUFF LENGTH

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Abstract: the effect of finite cuff length on the signals recorded by electrodes at different positions along the nerve was analysed in a model study. Relations were derived using a one-dimensional model. These were evaluated in a more realistic axially symmetric 3D model. This evaluation indicated that the cuff appeared shorter because of edge effects at the beginning and end of the cuff. The method for velocity selective filtering introduced by Donaldson was subsequently analysed. In this method, velocity selective filtering is achieved by summing the signals of subsequent tripoles after applying time shifts tuned to a certain conduction velocity. It was also found that the optimum electrode distance for a given cuff length for maximum summed RMS of symmetrical tripoles in the cuff is larger than when evaluating peak-peak amplitudes of single fibre action potentials. Velocity selective filtering yields better selectivity when using symmetrical tripoles, but may yield larger signal RMS when using the wider asymmetrical tripoles, potentially allowing for shorter cuffs. It is speculated that application of a multi-electrode reference may improve velocity selectivity for asymmetrical tripoles.

Introduction

Nerves transport bidirectional information between the central nervous system and the periphery. The conduction velocities of the action potentials travelling along the nerve fibers depend on the diameter of the fibers. Derivation of this bidirectional information transfer can be important in neural prostheses and other applications.

Donaldson et al. [1] proposed a method for velocity and direction selective recording using multi-electrode cuffs. This selectivity is achieved by adding delayed signals from subsequent electrodes, the delays being matched to a certain propagation velocity. This principle has recently been demonstrated experimentally [2].

The goal of the current study is to investigate the effect of finite cuff length on the signals recorded by the electrodes at different positions along the nerve. The effects may be important in optimising the velocity-selective recording of these signals.

Analysis

Consider a cylindrical cuff with zero conductivity around a nerve trunk, as indicated in figure 1. We assume axial-symmetry of the cuff and nerve. The cuff has length L and inner diameter D_i, resulting in an internal surface area $A_i = \pi(D_i/2)^2$.

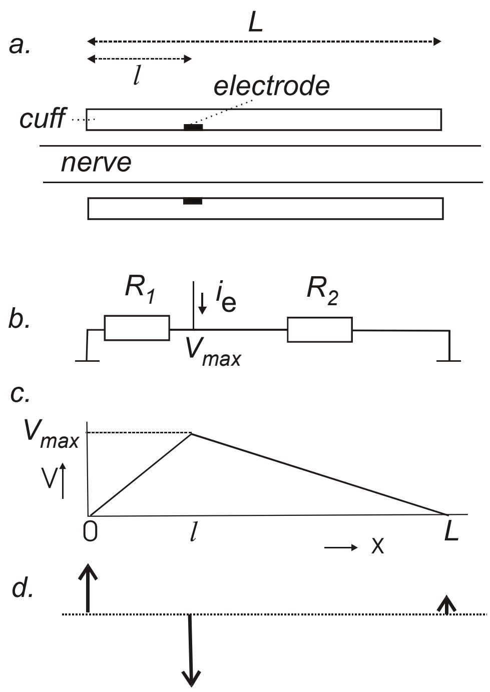

Figure 1: Derivation of weighting function for an electrode at position l along a cuff of length L. a: cross section of cylindrical cuff configuration with circular electrode on position l. b: effective resistive loading of electrode at position l if a current i were injected at this electrode. c: resulting weighting function $W_l(x)$, and d: its second spatial derivative.

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The potential $V_l(t)$ measured at any time t on an electrode at position l resulting from an action potential of a nerve fibre in the nerve trunk can be found by convolution of a weighting function $W_l(x)$ and the trans-membrane current density per unit length $j_m(x,t)$ [3]:

$$V_l(t) = \int W_l(x) j_m(x,t)dx \qquad (1)$$

The weighting function can be found by application of the reciprocity theorem, assuming the injection of a current $i_e = \int_{L_e} j_e dx$ at the electrode (figure 1). If the length of the electrode is relatively small with respect to the other longitudinal dimensions of the model, we can assume that the current is injected at one position, with current i_e being proportional to the average current density $\bar{j}_e$ over the length of the electrode $i_e = L_e \bar{j}_e$. If assuming a one dimensional conduction within the cuff and a single specific longitudinal conductivity σ_l of the tissue within the cuff, $W_l(x)$ follows (figure 1c):

$$W_l(x) = W_{l,\max} f_l(x) \qquad (2)$$

With $f_l(x) = \dfrac{x}{l}$ 0<x<l; $f_l(x) = \dfrac{L-x}{L-l}$ l<x<L (3)

The maxima of the weighting functions share a parabolic function of electrode position l with a maximum at $l=L/2$ (figure 2):

$$W_{l,\max} = \frac{L_e}{\sigma_l A_i} \frac{l(L-l)}{L} \qquad (4)$$

The transmembrane current density per unit length $j_m(x,t)$ is proportional to the second derivative with respect to position of the membrane potential $V_m(x,t)$:

$$j_m(x,t) = \frac{1}{R_i} \frac{d^2 V_m(x,t)}{dx^2} \qquad (5)$$

R_i being the intracellular resistance per unit length. Therefore, the electrode potential at time t can also be calculated as follows (see (1) and (5)):

$$V_l(t) = \frac{1}{R_i} \int W_l(x) \frac{d^2 V_m(x,t)}{dx^2} dx \qquad (6)$$

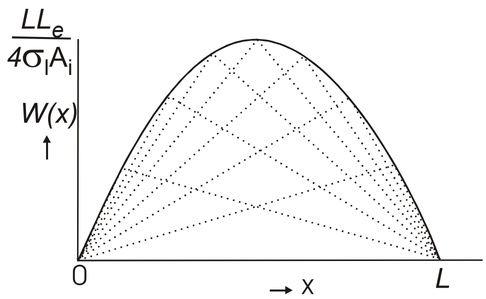

Figure 2: Weighting function W(x) for 7 equally spaced electrodes along a cuff of length L. The maxima of these functions constitute a parabolic function of electrode position l with a maximum at $l=L/2$.

Applying basic mathematical rules for differentiation and integration, and assuming that the weighting function $W_l(x)$ and its first spatial derivative tend to zero for $x \to \infty$ and $x \to -\infty$, this can be rewritten as [4]:

$$V_l(t) = \sigma_i \int \frac{d^2 W_l(x)}{dx^2} V_m(x,t)dx \qquad (7)$$

Thus, the electrode potential as a function of time can be derived by convolution of the second spatial derivative of the weighting function and the membrane potential.

The second derivative of the weighting function (2) consists of three delta functions (see figure 1d):

$$\frac{d^2 W_l(x)}{dx^2} = \frac{L_e}{\sigma_l A_i}\left(\delta(x)\frac{L-l}{L} - \delta(x-l) + \delta(x-L)\frac{l}{L}\right) \qquad (8)$$

Therefore, the signal measured monopolarly by an electrode at position l is triphasic, the first phase occurring when the action potential enters the cuff, the second when it is under the electrode and the third when it leaves the cuff (refer also to [5]). The action potential is weighted the same for the second phase independent of electrode position, the weighting varies with electrode position for the first and third phase. It should be noted that monopolar sensing against a relatively far reference electrode is not an option for ENG measurement, rather a tripolar configuration should be used in order to reject disturbances from electrical sources outside the cuff. We will now consider two configurations of tripolar electrodes:

- <u>Configuration 1:</u> A central electrode at position l with the other electrodes at the edge of the cuff (at $l=0$ and $l=L$)
- <u>Configuration 2</u>: A central electrode at position l with both other electrodes at a fixed distance Δ. This configuration is the configuration proposed by Donaldson et al [2].

<u>Configuration 1</u> has the same effective weighting function as the monopolar electrode, since the contribution of the edge electrodes to the weighting function is zero or small (see figure 2). If the edge electrodes are not at the very edge of the cuff, but more inside, the cuff appears shorter, having a length equal to the distance between the edge electrodes.

<u>Configuration 2</u> has a symmetrical weighting function, with the following second position derivative:

$$\frac{d^2 W_l(x)}{dx^2} = \frac{\Delta L_e}{\sigma_l A_i}\left(\delta(x-l-\Delta)\frac{1}{2} - \delta(x-l) + \delta(x-l+\Delta)\frac{1}{2}\right) \qquad (9)$$

It should be noted that the above analysis assumes a spatially continuous nerve fibre, which is realistic for non-myelinated nerve fibres. A similar spatially discrete analysis can be made for myelinated nerve fibres with spatially discrete nodes of Ranvier [3,4].

Donaldson et al [1], Rieger et al. [2] and Winter et al. [5] have presented and analysed diameter selective filtering using a multi-electrode configuration with

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equal distance Δ between the electrodes, and shifting the signals of subsequent tripoles in time over a delay tuned to the conduction velocity of the desired fiber diameter and direction of signal transfer. Thus, in a multi-electrode configuration of N electrodes, N-2 tripoles are available (see figure 3a). If the same electrode configuration is used for selective recording of another fibre diameter, the optimal inter-electrode distance for maximal tripole signal is proportional to fibre diameter [4]. However, for a given cuff-length, this results in fewer tripoles (figure 3b), which reduces summed signal strength of the delayed tripole signals [5]. Winter et al [5] therefore concludes that the optimum electrode distance for a given cuff length is approximately 3 mm when evaluating the peak to peak amplitude of single fibre action potentials. This result is independent of fibre diameter and assuming the same configuration of tripoles (figure 3a,b). It should be noted that signal to noise ratio should be optimised and that the RMS value of noise can be assumed to be proportional to the square root of the number of tripoles, while the RMS of the signal is proportional to the number of tripoles, assuming unchanged inter-electrode distances [5].

In the case of a given electrode configuration, with a certain number of electrodes *N*, and a certain cuff length *L,* and therefore a fixed inter-electrode distance $\Delta = L/(N-1)$, other combinations of electrodes are possible to form tripolar configurations (figure 3c,d).

We will now investigate whether this may potentially lead to an improved discrimination between signals from different diameter fibres and higher RMS value of the summed signals.

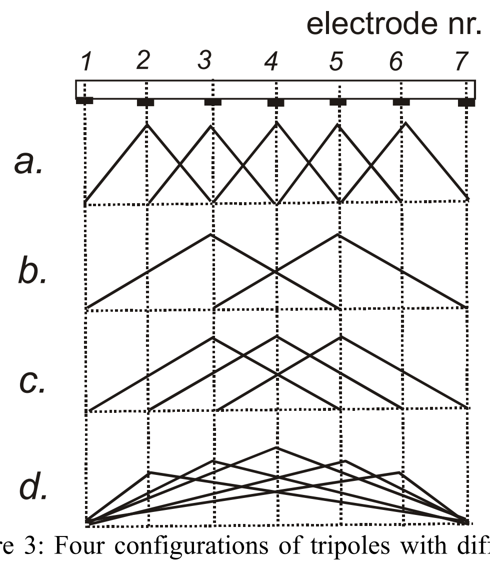

Figure 3: Four configurations of tripoles with different electrode distances, optimising for different diameter nerve fibers relative to cuff length.

Methods

A 3-D multi-electrode cuff electrode was modelled in 2D assuming axial symmetry [3]. The inner diameter of the cuff was 2 mm, cuff length was 30 or 60 mm, cuff conductivity was very low (10^{-6} $(\Omega m)^{-1}$). The nerve was modelled to have an outer radius of 0.55 mm and an inner radius of 0.5 mm. In between the epineurium was modelled, having a conductivity of 0.0034 $(\Omega m)^{-1}$. The nerve fascicle modelled inside the epineurium had an anisotropic conductivity of 0.6 $(\Omega m)^{-1}$ in the longitudinal direction and 0.083 $(\Omega m)^{-1}$ perpendicular to this direction. The medium between nerve and cuff and outside the cuff was modelled to be homogeneous and having a conductivity of 1 $(\Omega m)^{-1}$. The model was implemented in FEMLAB. Myelinated nerve fibres were modelled on the axis of the model, according to Wesselink et al [6], however, the relations between fibre diameter and internodal distance and between inner and outer diameter of the myelinated fibres were linearized in order to be able to simulate small diameter fibres as well as larger diameter fibers. In this paper, only fibres with an effective outer diameter of 5 μm and 10 μm are considered.

Results

The parabolic function connecting the maxima of the weighting functions of the asymmetrical tripoles of the multi-electrode configuration is deformed at the edges (figure 4), resulting in an effectively shorter cuff and increased weighting functions of symmetrical tripoles (configuration 3a) at the edges of the cuff (figure 5).

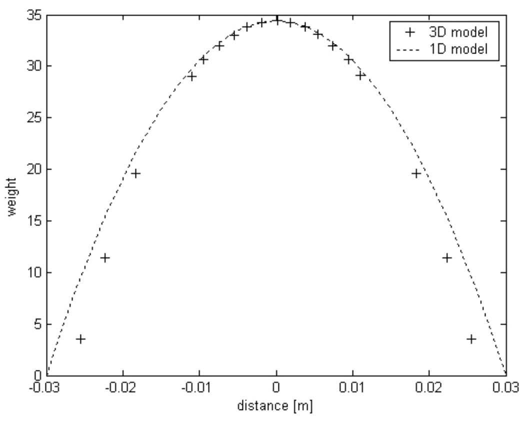

Figure 4: Influence of finite cuff length on the function connecting the maxima of the weighting functions of the assymetrical tripoles introduced in figure 2 and 3d. Cuff length was 60 mm.

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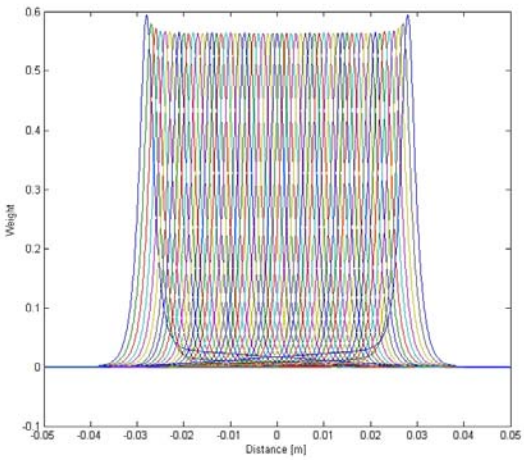

Figure 5: Weighting functions of symmetrical tripolar configurations formed of electrodes of a multi-electrode configuration of length 60 mm, electrode distance is 4 mm. Note the increased weighting functions at the edges due to the finite length of the cuff.

The optimal electrode distance for tripoles of configuration figure 3a for a given cuff length depends on fibre diameter, when evaluating RMS voltage in stead of peak-to-peak voltage as in [5]. RMS value gives a better indication of the contribution of the signal to a compound ENG signal than peak-to-peak voltage [4]. For 5 µm diameter fibres, the optimum electrode spacing for maximizing summed RMS signal was 5 mm in stead of 3 mm found on when considering basis of peak-to-peak amplitude [5]. When considering configuration c of figure 3 with an electrode distance of 1 mm, the optimal distance between the electrodes in each tripole when maximizing RMS signal was between 7 and 8 mm.

The signals of configurations a, c and d of figure 3 were evaluated with 5 and 10 µm diameter fibres. Cuff length was 30 mm, number of electrodes was 7, inter-electrode distance was 5 mm. The RMS value of the summed signals after optimal time shifting indeed increased for the 10 µm diameter fibre when going from configuration *a* to *c* and *d* by a factor 1.3 and 2.0 respectively, while this increase was, as expected from the above analysis, not as large for the 5 µm diameter fibre (a factor 1.0 and 1.4 respectively). However, the velocity-selectivity of configuration *a* was best for both the 5 and 10 µm diameter fibres.

Velocity selective filter tuned for 5 µm fibres			
	5 µm	10 µm	Relative
Conf. a	7.7	4,2	1.8
Conf. c	8.0	13,9	0.6
Conf. d	11.0	12,0	0.9
Velocity selective filter tuned for 10 µm fibres			
	5 µm	10 µm	relative
Conf. a	3.2	16,2	5.1
Conf. c	6.8	21,2	3.1
Conf. d	11.1	33,6	3.0

Table 1: RMS of summed signals of a single fibre after time shifting tuned to myelinated nerve fibres having an effective diameters of 5 or 10 µm. The RMS errors were determined over a 5 ms period and are expressed in nV. Cuff length was 30 mm.

Discussion

The weighting functions of asymmetric tripoles with first and third electrode placed at the beginning and end of the cuff have the maximum effective width, but shape of the signal is varying. Asymmetric tripoles indeed yielded largest RMS value, especially for signals of larger diameter fibres, however, velocity selectivity using a velocity matched filtering was not optimal, since the first and last phases varied depending on the position of the middle electrode. This may be improved by evaluating the signals on each electrode relative to the mean value of the signals of all electrodes (multi-reference: figure 6). This procedure is commonly used in multi-electrode EEG signal analysis. The multi-reference signals only have one major phase, the first and third phase are relatively small (figure 7). However, there is an offset on the signal, since the time-average must be zero (figure 7). This could be rejected by using a suitable high-pass filter before applying the velocity-matched filtering.

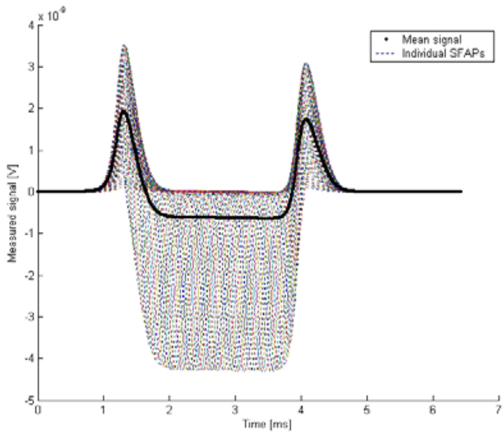

Figure 6: Mean signal of all electrodes, which can be used as a multi-reference to the individual signals. Cuff length was 60 mm, inter-electrode distances was 4 mm.

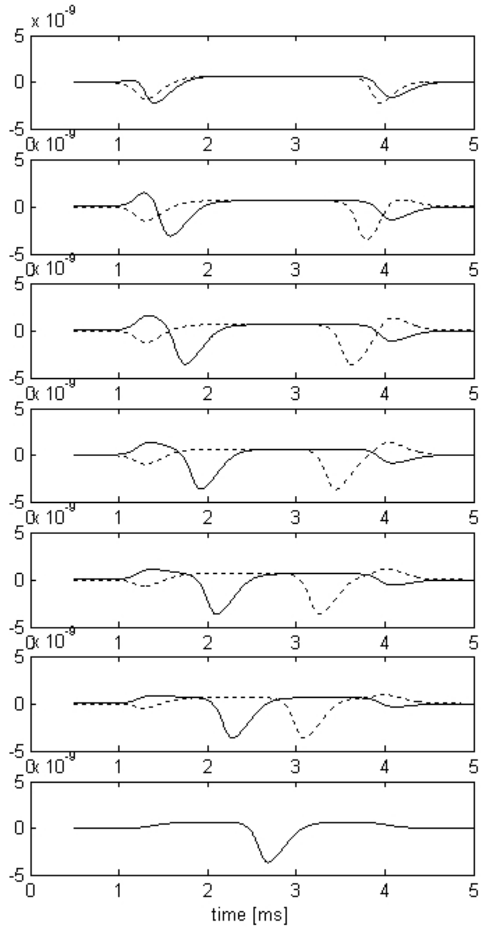

Figure 7: Multi-electrode signals relative against the mean signal of all electrodes (multi-reference). Cuff length was 60 mm, inter-electrode distance 4 mm.

Asymmetric tripole configurations may only be of interest for selective sensing of relatively thick fibers. However, it also means that for a given maximal nerve fiber diameter of interest, cuff length can be minimal.

Conclusions

1. A cuff of finite length appears electrically shorter than its actual length
2. Optimum electrode distance for given cuff length is larger when evaluating signal RMS than peak-peak amplitude.
3. Velocity-selective filtering using a multi-electrode configuration yields better velocity sensitivity when using symmetrical tripoles than when using asymmetrical tripoles with first and last electrode at the beginning and end of the cuff, but the RMS of the summed signals is lower.

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