ARTICLE IN PRESS
JID: JJBE [m5G;December 9, 2016;21:16]

Medical Engineering and Physics 000 (2016) 1–8

Contents lists available at ScienceDirect

Medical Engineering and Physics

journal homepage: www.elsevier.com/locate/medengphy

Characterizing the reduction of stimulation artifact noise in a tripolar nerve cuff electrode by application of a conductive shield layer

Parisa Sabetian[a], Bita Sadeghlo[b], Chengran Harvey Zhang[b], Paul B Yoo[a,b,*]

[a] Institute of Biomaterials and Biomedical Engineering, University of Toronto, 164 College St, Room 407, M5S 3G9, Toronto, ON, Canada
[b] Department of Electrical and Computer Engineering, University of Toronto, 10 King's College Road, Room SFB540, M5S 3G4, Toronto, ON Canada

ARTICLE INFO

Article history:
Received 3 February 2016
Revised 14 November 2016
Accepted 27 November 2016
Available online xxx

Keywords:
Peripheral nerve recording
Artifact signal
Single fiber
Action potential
Finite element model
Neuroprosthesis

ABSTRACT

Tripolar nerve cuff electrodes have been widely used for measuring peripheral nerve activity. However, despite the high signal-to-noise ratio levels that can be achieved with this recording configuration, the clinical use of cuff electrodes in closed-loop controlled neuroprostheses remains limited. This is largely attributed to artifact noise signals that contaminate the recorded neural activity. In this study, we investigated the use of a conductive shield layer (CSL) as a means of reducing the artifact noise recorded by nerve cuff electrodes. Using both computational simulations and in vivo experiments, we found that the CSL can result in up to an 85% decrease in the recorded artifact signal. Both the electrical conductivity and the surface area of the CSL were identified as important design criteria. Although this study shows that the CSL can significantly reduce artifact noise in tripolar nerve cuff electrodes, long-term implant studies are needed to validate our findings.

1. Introduction

Peripheral nerve cuff electrodes provide a robust platform for implementing neurostimulation therapies. As one of the most extensively studied type of neural interface, cuff electrodes have been used successfully in myriad clinical applications that involve electrical nerve stimulation: bladder dysfunction [1,2], depression and epilepsy [3,4], respiratory disorders [5–7], and functional reanimation of upper and lower extremities [8,9]. In parallel to these efforts, investigators have also demonstrated the feasibility of using peripheral nerve activity as a potential means of optimizing therapies via closed-loop controlled nerve stimulation. Examples include detection of bladder activity by pelvic nerve recording [10], estimation of ictal or inter-ictal activity via the vagus nerve [11], prediction of airway obstruction via the hypoglossal nerve [12,13], and gait detection by sural nerve recording [14,15].

However, the wide spread clinical use of nerve electrodes as a means of recording peripheral nerve activity has been limited [16]. This is due to myriad factors such as (1) the inherently poor signal to noise ratio achieved by nerve cuff electrodes [17,18], (2) clinical translation of advanced neural interfaces (e.g., intrafascicular electrodes) for long-term use in patients [19], and (3) development of closed-loop controllers that can process and utilize the recorded neural signals [20]. Even with the use of insulating material (e.g., silicone elastomer) to improve the recording of neural activity with nerve cuff electrodes, contamination of the electroneurogram (ENG) by artifact noise has proven unavoidable. Compared to the µV-level activity recorded from peripheral nerve fibers, artifact noise generated by external sources (e.g., muscle, movement artifacts, electrical interference) can contaminate the ENG with signals that (1) overlap with respect to frequency content and (2) are several orders of magnitude larger in amplitude [18,21].

The tripolar nerve cuff electrode design is considered the gold-standard for achieving low-noise measurement of peripheral nerve activity [18]. By differentially measuring the electric potentials between a single middle contact and the two electrically-shorted symmetrical side contacts, any voltage drop (i.e., external noise) that is generated along the inside of the nerve cuff is effectively eliminated. And while results from human studies support the feasibility for long-term implants [14,22], any changes or imbalances in the electrical resistivity along the nerve cuff (e.g., encapsulation [21], and imperfect electrode placement [15,23]) are known to undermine significantly the noise-cancelling effects of the tripolar configuration.

In this study, we investigated the feasibility of improving the recording properties of a tripolar nerve cuff electrode with a conductive shield layer (CSL) by quantifying changes in the recorded stimulus artifact. Using a computational model, we systematically characterized the effects of the CSL (e.g., physical dimensions) on

* Corresponding author at: Institute of Biomaterials and Biomedical Engineering, University of Toronto, 164 College St, Room 407, M5S 3G9, Toronto, ON, Canada. Fax: +1 416 978 4317.
E-mail address: paul.yoo@utoronto.ca (P.B. Yoo).

http://dx.doi.org/10.1016/j.medengphy.2016.11.010
1350-4533/

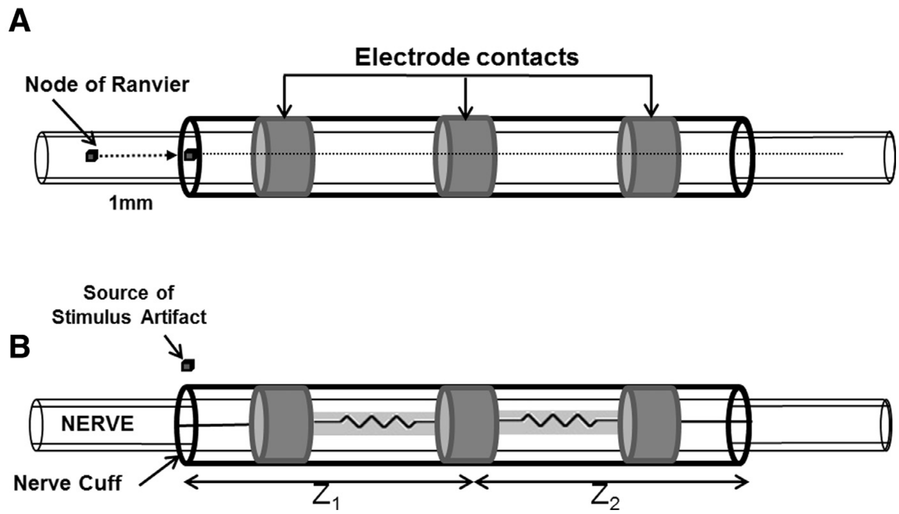

Fig. 1. The finite element model (FEM) of a peripheral nerve (PN) recording system. The impedance mismatch was computationally modeled, where the conductivity of Z_1 and Z_2 were varied such that the total resistance of cuff remained constant. The mismatch ratios varied from 1 to 100. (A) Movement of the electrical source inside the nerve was used to simulate the effects of saltatory action potential (AP) conduction. (B) The electrical-source was used to simulate artifact noise.

Table 1
Geometries and electrical properties of computer model components.

Parameters	Diameter (mm)	Length (mm)	Electrical conductivity (S/m)	Reference (s)
Epineurium	1	40	$8.26E^{-2}$	[12,43]
Endoneurium	0.8	40	$8.26E^{-2}$ (Transverse) $5.7E^{-1}$ (Longitudinal)	[12,43]
Perineurium	0.85	40	$2.09E^{-3}$	[12,43]
Nerve cuff (silicone)	1.25	13	$1.00E^{-17}$	[29]
Electrode contacts (platinum)	1.1	1	$9.44E^{+06}$	
Saline medium	30	60	1.05	
External shield layer (CSL)	1.3	13	$9.44E^{+06}$	
Current source-node of Ranvier ($I = 1$ nA)	N/A	0.001	$9.44E^{+06}$	

both the simulated single fiber action potentials (SFAP) and artifact noise generated by an electrical source located external to the nerve cuff. In-vivo animal experiments were also conducted to validate the noise-reducing effects of the CSL that were demonstrated in our computer model. Preliminary results of this study were presented as a conference paper [24].

2. Materials and methods

2.1. Computational study

2.1.1. Simulation of single fiber nerve action potential (SFAP)

As described in a previous study [12], a two-step computational method was used to simulate SFAPs recorded by a tripolar nerve cuff electrode implanted around a mono-fascicular nerve (Fig. 1). The first step involved constructing a volume conductor model using finite element software (Comsol Multiphysics, Comsol Inc.). The nerve consisted of an endoneurium with concentric layers of perineurium (thickness = 25 μm) and epineurium (thickness = 75 μm). At the mid-point along the nerve, a tripolar nerve cuff electrode was modeled as an electrically-insulating silicone tube (thickness = 0.125 mm) with three platinum ring electrodes embedded along the inner surface of the cuff (Fig. 2). A thin layer of saline (thickness = 25 μm) was also added between the nerve and the inner surface of the cuff. A single current source located within the center of the endoneurium was used to approximate a node of Ranvier, while the outer surface of the large saline bath was set as the electrical ground. The physical properties of the model are summarized in Table 1. The model was used to solve the electric potentials at each of the three platinum contacts that were generated by a single node of Ranvier (current source = 1 nA). Simulations were repeated as the node of Ranvier was displaced in 1 mm increments along the length of the nerve (i.e., single 10 μm myelinated axon, Fig. 1(A)), and thereby provided a complete spatially-dependent array of voltages measured at each electrode contact.

The second step of generating SFAPs involved a MATLAB coded algorithm (Mathworks Inc., Natick MA) that implemented time-dependent changes in the electrode contact voltages. The trans-membrane current of a node of Ranvier obtained from Neuron software was used to scale the electrode contact voltages; while both the saltatory conduction velocity of a 10 μm fiber (inter-nodal delay = 17.9 µs) and the duration of membrane current was used to determine the number of active nodes of Ranvier for any given time. The resulting tripolar nerve cuff recordings were calculated by the following equation:

$$SFAP = V_{\mathrm{mid}} - V_{\mathrm{side,avg}} \quad (1)$$

where, V_{mid} is the voltage at the middle electrode contact, and $V_{\mathrm{side,avg}}$ is the average voltage of the two side electrode contacts.

2.1.2. Stimulus artifact

The model was used to generate a stimulus artifact signal by moving the current source (node of Ranvier in previous section, Fig. 1(B)) outside the nerve cuff electrode: 3 mm from the center of the nerve trunk and at different longitudinal locations from the mid-point of the tripolar nerve cuff (5 mm, 10 mm, and 15 mm). Since an ideal tripolar nerve electrode does not generate a stimulus artifact, we introduced an imbalance in the electrical resistance along the nerve cuff (Z_1 and Z_2, Fig. 1(B)), which approximated biological changes at the nerve–electrode interface in chronically implanted devices. With the electrical conductivity of Z_1 maintained

Please cite this article as: P. Sabetian et al., Characterizing the reduction of stimulation artifact noise in a tripolar nerve cuff electrode by application of a conductive shield layer, Medical Engineering and Physics (2016), http://dx.doi.org/10.1016/j.medengphy.2016.11.010