EXTRACTION OF SOMATOSENSORY INFORMATION FROM PERIPHERAL NERVES FOR FNS APPLICATIONS

Schoenberg A. A., Malagodi M., Horch K.
Bioengineering Department, University of Utah, Salt Lake City, USA

Abstract

We are developing electrodes for chronic recording of activity in tactile nerve fibers of peripheral nerves for use in feedback control of FNS systems. The new method involves threading 25µm diameter wires longitudinally inside individual fascicles of peripheral nerves. The electrodes were made from insulated Platinum wires in which 0.25 - 4.0 mm of insulation was removed. The electrodes were then attached to a 50µm diameter Tungsten needle that was stiff enough to allow threading it through the nerve. Acute experiments have been conducted in rat sciatic nerves. The signal-to-noise ratios of the sensory fiber recordings were measured as a function of the amount of uninsulated length **of the platinum** wire. Both monopolar and bipolar recordings were made. The best signal-to-noise ratio was obtained with tip exposures between 0.5 and 2.0 mm. The average signal to noise ratio for these length was approximately 2. Spectral analysis of the recordings showed that most of the signal power appears in the octave band between 750 and 1500 Hz. The bipolar recordings **were** less susceptible to power line frequency interference.

key words: Electrodes, nerve recording, somatosensory, neuroprostheses, FNS

Introduction

The research is motivated by the need to provide better rehabilitation for people with somatosensory loss and paralysis caused by spinal cord injury. Spinal cord injuries occur at an estimated rate of 50 per one million population per year [Kalsbeek et al., 1980]. This translates into 10,000 new cases per year in the U.S. alone. U.S. Hospitals treated 28,000 spinal cord injury cases in 1974 at a cost of 230 million 1974 dollars [Krauss, 1980]. Very few patients with complete spinal cord injury regain any useful motor or sensory function [Dimitrijevic, 1983]. These figures provide ample justification for developing prostheses or therapy that would allow patients to perform unassisted activities of daily living such as personal hygiene, using the telephone, feeding and clothing themselves.

<u>Neural Prostheses:</u> In the near term, neural prostheses of various types show promise as a method of restoring lost function [Hambrecht, 1979, 1982]. For selected quadriplegics the possibility of restoring key and pinch grasp using transcutaneous stimulating electrodes has been demonstrated [Peckham et al., 1980]. Functional neuro-muscular stimulation (FNS) of lower extremities to enhance crutch assisted walking in paraplegics has been advanced with application of external electrodes [Kralj et al., 1983; Petrofsky et al., 1984; Popovic et al., 1986] as well as with implanted electrodes [Marsolais, 1986; Stoehr et al., 1983]. Before present techniques of neuromuscular stimulation can move out of the experimental laboratory to a clinically acceptable therapy, a number of significant improvements are needed: (1) reliable implantable electronics and electrodes; (2) more reliable and larger numbers of

command channels of volitional control; and (3) provision for somatosensory feedback.

The present study addresses primarily the third need - to provide tactile feedback to, the paraplegic or quadriplegic patient via remaining intact sensory pathways as well as to provide such feedback to the next generation of electronic stimulation controllers. The direct approach of developing artificial sensory transducers that mount on the paralyzed hand has been under development at the University of Utah [Schoenberg et al., 1984] as well as at other centers [Mortimer et al., 1983; Petrofsky et al., 1984]. Externally mounted transducers, particularly those incorporating multiple sensing elements, require numerous wires and complex electronics. They are not likely to soon attain the reliability and ruggedness required for mounting on the external grasping surfaces of the hand. Furthermore, externally mounted transducers are not likely to be used by paralyzed patients due to the difficulty of attachment and removal, and the strong psychological aversion to devices that emphasize or display the quadriplegic's disability.

The ideal device is implantable, multichannel and reliable. These criteria are met by the physiologic sensors of tactile force and movement, that are still functioning in the fingers and hand of the quadriplegic. The tactile information is available in the form of action potentials traveling along multiple parallel axons of the peripheral nerves. It is the interruption of the ascending pathways at the site of the spinal cord lesion that prevents the quadriplegic from obtaining this useful somatosensory information. If extracted from the peripheral nerves via small implantable electrodes or sensors, such information can be amplified and used to stimulate alternate sensory pathways that are not affected by the spinal lesion. External cutaneous stimulation is currently being developed to bypass the lesion [Solomonow and Lyman, 1980; Szeto and Saunders, 1982]. However, direct stimulation of afferent nerves is in the realm of possibility [Dimitrijevic, 1983; Stein et al., 1980]. A more immediate use of tactile information such as finger pressure would be as a direct electrical feedback signal to control FNS of paralyzed muscle.

Chronic Neural Recording: A number of techniques show promise for chronic recording from peripheral nerves. Stein et al. (1975) reported on two types of recording techniques. One is the so called "regeneration electrode", which consists of multiple electrodes imbedded in a porous substrate. The porous channels allow regenerating axons of a transected nerve to grow into them and thereby come into intimate contact with the imbedded electrodes. This worked with the sciatic nerve of amphibians but proved unsuccessful with mammalian nerves. This technique is not likely to be applicable to sensory nerve recording since it requires transection of the nerve thereby causing the denervation of the tactile sensors. A second approach is to use a silastic cuff containing a tripolar electrode placed around the nerve [Hoffer and Loeb, 1980; Loeb et al., 1977; Stein et al., 1975]. The central electrode is connected to one side of a differential amplifier and the two other electrodes, placed near each end of the cuff, are connected together and attached to the other side of the differential amplifier. The advantage of this configuration is that interfering EMG signals and other common mode noise sources are greatly suppressed. The insulating cuff also provides a high impedance path for the small nerve currents, thus increasing the recorded voltages. The cuff method records a weighted average of the active fibers in the whole nerve and hence is not likely to be useful where a significant mix of fibers from various locations is present. A difficulty with the cuff is that it must be correctly sized for the nerve so that it does not impair the nerve's blood supply.

Loeb and coworkers [Hoffer and Loeb, 1980; Loeb et al., 1977] have described a new technique for long term unit recording of somatosensory neurons in the dorsal root ganglia of cats. This technique has since then been improved to allow recordings of single units in the dorsal root ganglia of up to 2 months [Hoffer and Loeb 1980; Hoffer et al., 1981]. The technique involves insertion of 25 µm diameter platinum-iridium "hatpin" electrodes welded to a flexible gold lead. The weld is protected with an epoxy bead and the whole assembly is insulated with Parylene. Electrode life is limited due to breakage of the leadout wires, local tissue damage, and electrode tip encapsulation which reduces the recording amplitude. These authors suggest that the current version of the electrodes are not suitable for prostheses applications where much longer and more stable recording would be required.

A new technique for chronic recording of single unit activity in small peripheral nerves has been described by a group in Belgium [Janssens et al., 1979]. A 27.5 µm diameter wire was threaded through a small hole in the perineurium of the nerve and pushed along the inside of the fascicle for several centimeters. Unit activity of up to 300 µV from well defined receptive fields of the ankle was recorded. To eliminate the EMG signals which obscured the unit sensory activity during muscle activation, the method was modified to bipolar recording. Two wires were glued together with their exposed tips separated by 1 mm. The two wires were connected via identical buffer amplifiers to a differential amplifier. The recorded action potentials of 200 µV were well above the background activity of the nerve and exceeded the interfering EMG by a factor of 4. The paper implied that long term recording was possible with this technique. However, the data shown extend only to 24 hours after implantation.

The approach of inserting flexible wire electrodes inside fascicular subunits of the nerve in close contact with the axons has the appeal of simplicity and the possibility of multichannel sensory signal recording. In the first stage of our study the goal has been to develop a reliable method for inserting electrodes into fascicles and determining the configuration which would yield the highest signal and minimize the noise.

Methods

The experiments used the sciatic nerves of adult rats as a model since the two fascicles of this nerve at mid-thigh level are similar in size and mechanical properties to fascicles in human sensory nerves.

Studies of electrode configuration to date have involved two variables: the length of the exposed (recording) zone and the separation between the members of the bipolar pairs. Parametric studies of signal-to-noise ratio, as described below, have been done for various exposed lengths and bipolar separations of 25 µm diameter Platinum (Pt) electrodes.

The Pt electrodes were fabricated from Teflon or Isonel insulated wires. The desired amount of insulation was removed by protecting the surrounding wire by a heat sink and heating the target area with a small flame. Exposed lengths ranged from 0.25 to 4.0 mm.

Electrode impedance was measured as a function of exposed electrode length with a 1000 Hz sinusoidal current. The results are plotted in Figure 1. Note that zero length exposure represents the impedance of the insulated wire when only the cut end of the wire was exposed. For the 25 µm wire this impedance was in the range of 1 Mohm. The impedance drops by a factor of 20 to approximately 50 kohms at 0.5 mm exposure, and decreases further. to 10 kohms at 2.0 mm exposure. This data allowed us to determine exposure length by measuring electrode impedance.

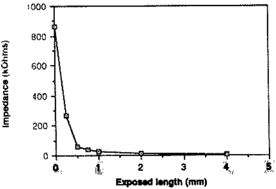

Figure 1. Electrode impedance as a function of recording zone length.

The platinum wires are too flexible to be pushed into the nerve. Hence we have developed a method of inserting the electrode into the fascicle involving a sharpened tungsten needle. The configuration of the insertion assembly is shown in Figure 2. A 50 µm tungsten wire is electrochemically sharpened at one end and beveled at the other end to form a smaller diameter attachment and a protective shoulder as shown. The tungsten needle has enough mechanical stiffness to be threaded longitudinally into a fascicle. The electrode is then attached to the beveled end of the tungsten needle with cyanoacrylate glue. Alternatively, a 25 µm diameter lead wire is glued to the beveled end of the Tungsten needle and twisted together with the Pt electrode as shown in Figure 2.

These electrodes were inserted into the nerve as follows. The sciatic nerve of the anesthetized rat was exposed at mid-thigh level, and its two fascicles were separated by blunt dissection over a distance of about 1 cm. Under 25 x magnification, the tungsten needle was pushed through the perineurium and threaded longitudinally through the exposed fascicle. The needle exited the fascicle approximately 1 cm distal to its entry point and was pulled through until the recording zone of the electrode was approximately midway between the entry and exit points. It was necessary to expose individual fascicles because threading the needle in the whole nerve rarely produced strong signals since the electrode seldom end up inside a fascicle.

Once the electrode was inside the fascicle, the lead wire was cut near to where it exited the perineurium. The incision was then closed, and a needle electrode was inserted percutaneously to serve as the reference or ground electrode. Similar techniques were used for the bipolar recordings, except that two electrode wires were wound around the lead wire.

The electrodes were connected to a high impedance amplifier with a frequency response of 100 Hz to 8000 Hz.

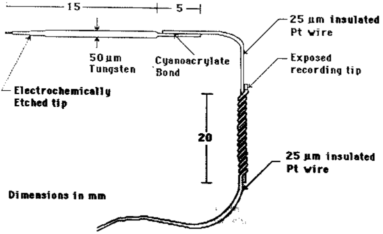

Figure 2. Schematic diagram of the micro-needle used to insert electrode into nerve fascicles.

Nerve activity was elicited by mechanical stimuli such as brushing the skin or squeezing the foot. Signal-to-noise ratios were quantitated in two ways. First, the peak-to-peak amplitude of the signal present during stimulation was divided by the peak-to-peak amplitude of the signal present in the absence of mechanical stimulation of the foot. For this method, a value of 1 means that there was no measurable change in the signal when the foot was stimulated, and a value of two means that the amplitude of the combined noise and signal was double that present without stimulation. A second method of measuring nerve activity was to count suprathreshold pulses during stimulation. The threshold level was set so that the pulse counter measured between 0 and 5 pulses per second during baseline conditions without any mechanical stimulation. The response to stimulation was then measured as the number of threshold crossings per second when the paw was mechanically stimulated.

Results

A typical oscilloscope trace of sciatic nerve activity for periodic mechanical squeezing of the paw is shown in Figure 3. In this particular case the recording was from the larger fascicle of the nerve with a electrode tip exposure of 1.0 mm. The signal-to-noise ratio for this recording was estimated to be 2.2. The amplitudes for noise and signal were estimated visually at a level where half of the pulses exceeded the selected amplitude. Similar recordings were obtained for 0.25, 0.5, 0.75, 1.0, 2.0, and 4.0 mm exposures of the recording zone. The data are shown in Figures 4 and 5. Figure 4 plots the signal-to-noise ratio as a function of electrode exposure. The largest signal-to-noise ratios were obtained with exposures in the range of 0.5 to 2.0 mm.

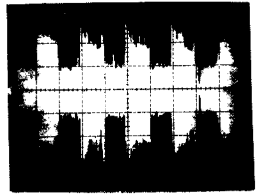

Figure 3. Recording of sciatic nerve sensory fiber activity using a monopolar 25 µm Pt electrode with 1.0 mm tip exposure. The horizontal scale is 1 s/div and the vertical scale is 50 µV/div. The paw of the rat was squeezed vigorously for 1 second intervals.

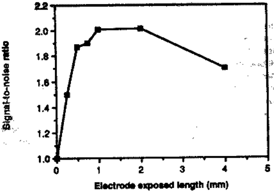

Figure 4. Signal-to-noise ratio vs exposed length of electrode showing averages of 10 trials for 25 µm diameter electrodes implanted in rat sciatic nerve fascicles.

Figure 5 shows the number of pulses in a second that exceeded a preset threshold vs exposed length that were recorded when the paw was squeezed. The average is plotted as a connected line. This data indicates that there is not much to be gained from increasing recording length beyond 1.0 mm but that lengths less than 0.5 mm significantly reduce the signal. Similar data was obtained with a 50 μm diameter Pt wire. The signal-to-noise ratio for the larger wire was not significantly different from that obtained with 25 μm wire.

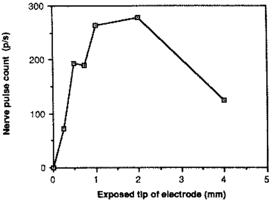

Figure 5. Pulse Counts per second as a function of exposed **recording length of a 25 μm diameter** Pt electrode showing **averages for 10 trials at each** length for rat sciatic nerve.

A more refined measure of signal-to-noise ratio was obtained by performing a power density spectral analysis of the recorded signals in the absence and presence of external stimulation. An example of such a frequency spectrum is shown in Figure 6. The middle curve, representing the difference between the stimulated and unstimulated conditions, shows the distribution of energy for the signals we are interested in extracting; namely, activity evoked by mechanical stimulation of the receptive fields of the sensory axons within the recording zone of the implanted electrode. Analysis of such data indicates that the octave band between 750 and 1500 Hz is a particularly important carrier of information. Therefore, we may increase our signal-to-noise ratios by filtering the recordings with an octave band-pass filter centered in this range.

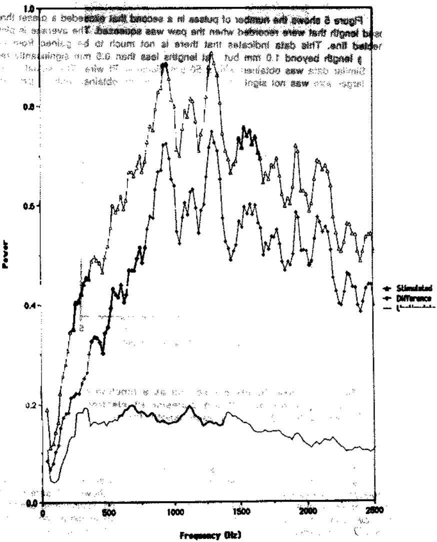

Figure 6. Power spectrum of sensory nerve activity for stimulated (upper) and unstimulated (lower)paw showing difference in spectra (middle). The difference spectrum peaks in the 750 to 1500 Hz range.

Discussion and Conclusions

The electrode insertion technique for the acute experiments has been developed to the stage where consistent recordings from nerve fascicles are achieved routinely. We conclude from the data that the best configuration for recording is to use electrode exposures in the range from 0.5 to 1.0 mm for both monopolar and bipolar recording. These recording areas also result in electrode impedances in the range of 15 to 30 kOhms.

The critical question that remains is how long intrafascicular recordings can be maintained. Preliminary tests in 2 cats show that recording amplitudes in fascicles of the ulnar nerve showed little or no diminution over a 48 hour period. This is an indication that the insertion of the electrodes does not produce excessive damage to the axons. Longer term tests of implanted electrodes are planned as soon as a reliable method is developed to anchor the electrodes to the nerve.

The electrodes implanted in the ulnar nerve of the cat were also tested for the voltage levels required to elicit contraction of muscles of the forearm. The intrafascicular electrodes, when pulsed with a 200 µs duration monophasic square wave, had threshold voltages in the range of 1.5 to 2.0 volts. Assuming an electrode impedance of 30 KOhms, the corresponding pulse current was estimated to be between 50 and 75 µA. These thresholds also changed very little over a 48 hour monitoring period.

Intrafascicular recording shows promise as a method of obtaining sensory information useful for feedback control of FNS, as well as providing a source of sensory input for patients with damaged central sensory pathways. A similar technique also shows promise as a totally implantable system for external control of muscle activation.

Acknowledgement

This work was supported by a grant from the National Institute of Handicapped Research (NIHR) of the U.S. Department of Education.

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