*Annals of Biomedical Engineering*, Vol. 19, pp. 197–207, 1991
Printed in the USA. 

0090-6964/91 $3.00 + .00


# Chronically Implanted Intrafascicular Recording Electrodes

Todd Lefurge, Eleanor Goodall, Kenneth Horch, Larry Stensaas, and Andrew Schoenberg

Departments of Bioengineering and Physiology
University of Utah
Salt Lake City, UT

*(Received 5/16/90; Revised 12/4/90)*

*A newly designed intrafascicular electrode for chronic neural recording was studied by implanting 12 electrodes in the radial nerves of 6 cats for 6 months. Action potentials were monitored at specified intervals throughout the experiment. The number and size of the signals recorded suggest that this type of electrode provides information that is appropriate for feedback control in functional electrical stimulation (FES) systems. Histology of the nerve revealed that the implants are biocompatible and that little damage is caused by the presence of the electrode.*

**Keywords**—*Neuroprosthetics, Peripheral nerve, Implanted electrodes.*

## INTRODUCTION

Functional Electrical Stimulation (FES) holds promise as a method for restoring motor function following spinal cord injury (2,10,16). However, sensory feedback is necessary to provide adequate control of FES systems. Feedback via afferent activity in peripheral nerves from proprioceptors and cutaneous mechanoreceptors could provide information which is more "functionally appropriate" than that provided by external devices (6).

The goal of this investigation was to determine whether an intrafascicular electrode developed in our laboratory (12,14) could be used to collect somatic sensory information over a period of several months. In particular, we wanted to know if activity could be recorded from a group of nerve fibers which was both representative of the entire fascicular population, yet small enough so that single units could be identified with a real-time spike separation system.

## MATERIALS AND METHODS

### *Electrode Fabrication*

Bipolar electrodes were constructed from Teflon insulated 90% Pt-10% Ir wires after the design of Malagodi *et al.* (12). A number of improvements in the electrodes

*Acknowledgments*—This research was supported by a grant from the National Institute of Disability and Rehabilitation Research. We thank Mike Todd for his assistance with the histology.

Address correspondence to Dr. K.W. Horch, Department Bioengineering, 2480 MEB, University of Utah, Salt Lake City, UT 84112.

were made during early phases of the study. We describe here the final and most successful version.

Recording sites were created by removing approximately 1 mm of insulation from the wires, at a distance of 2 cm from one end, with heat from a small platinum foil loop, and then depositing platinum black on the uninsulated region. A needle was fabricated from 50 $\mu$m diameter Tungsten wire by electrolytically etching it in a dilute solution of $KNO_2$, and attached to a 25 $\mu$m diameter Pt-Ir wire with cyanoacrylate glue.

Two wires, one with the needle attached and the other a 50 $\mu$m diameter reference electrode, were wrapped around a loop of 6-0 suture and the ensemble was threaded through Silastic tubing. A 1.5 cm-long tab was made at the end of the silastic tube to serve as an anchoring site. Medical grade Silastic adhesive was injected into the tubing to anchor the electrode wires and suture. After curing, the electrode assembly was washed with a mild soap solution, rinsed in distilled water, stored in a closed container, and autoclaved prior to implantation. Figure 1 shows the appearance of the finished electrode.

### *Animal Preparation and Electrode Implantation*

Adult cats were anesthetized with an intramuscular injection of Ketamine (33 mg/Kg) and maintained on a mixture of oxygen and Halothane. Rectal temperature was mon-

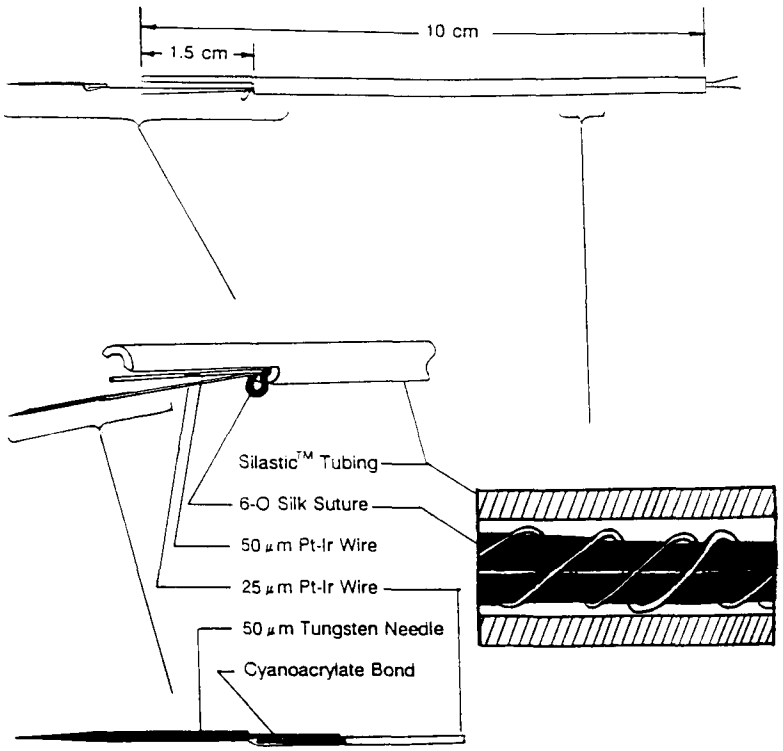


**FIGURE 1. Schematic drawing of the construction of an intrafascicular electrode. Although not to scale, the illustration shows the important features of the electrode design in enlargements.**

itored and kept at 37°C with a heated plate. Aseptic surgical procedures included removal of hair from the forelimbs, iodine sterilization of the skin, use of sterile drapes around the surgical field, and steam sterilization of all surgical equipment.

The radial nerve of each forelimb was exposed within a 5 cm incision, carefully separated from the surrounding connective tissue for about 3 cm, placed on a platform, and kept moist with sterile saline. The epineurium was carefully teased open under 25× magnification to expose the surface of one fascicle for approximately 2 cm.

The tungsten needle was used to thread the 25 $\mu$m wire inside the fascicle for about 1 cm, centering the recording area within this zone. The 50 $\mu$m reference electrode was placed outside the fascicle, with its recording area aligned to that of the intrafascicular electrode.

The 6-0 silk suture loop was secured to the epineurium proximal to the insertion site using 9-0 Ethilon suture; the Silastic tube was sutured to connective tissue approximately 1 cm proximal to the insertion site; the distal end of the wires and the tab of Silastic tubing were anchored to the epineurium; and the needle with its cyanoacrylate bond were cut from the wire. Figure 2 is a drawing of an implanted intrafascicular electrode.

After a second electrode pair was similarly implanted in another fascicle of the same radial nerve, the free ends of the electrodes were drawn through an incision above the elbow and the skin over the implantation zone closed. The free ends of the electrodes were capped, placed under the skin, and the incision closed.

Except where stated otherwise, the results presented here are based on 12 electrodes implanted in 6 cats for a period of 6 months.

## *Recording*

Recording sessions involving fully anesthetized animals took place immediately after implantation, and 0.5, 1, 2, 4, and 6 months thereafter. The cats were first elec-

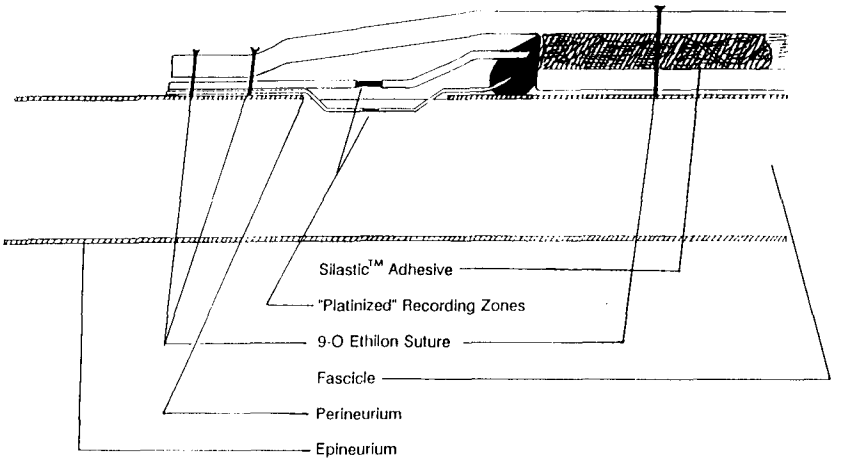


**FIGURE 2. Implanted intrafascicular electrode. Although not to scale, this figure illustrates the orientation of the electrode with respect to the fascicle and the attachment of the electrode to the nerve.**

trically grounded via a 25 gA subcutaneous needle above the elbow and the free ends of the electrodes retrieved through an incision above the elbow. Electrodes were capacitively coupled to a high impedance differential amplifier and band pass filter which had a gain of $1.5 \times 10^4$. Signals were displayed on an oscilloscope and recorded on magnetic tape.

Electrode impedance was measured, *in vivo*, using a 1 kHz constant current sine wave applied to each electrode. A subcutaneously placed hypodermic needle served as the return electrode in these measurements. Current was kept in the nano amp range to prevent any damage to the surrounding nerve fibers.

### *Identification of Single Units*

Individual units were selectively activated using controlled mechanical stimuli delivered to restricted regions of the skin and hair of the forepaw (11). Action potentials were displayed on an oscilloscope and the signal-to-noise ratio was measured as the peak-to-peak amplitude of the unit action potentials divided by the amplitude of the background signal when no stimulation occurred. If the signal-to-noise ratio for a single unit was greater than 1.4, the unit was included in the data set. The value 1.4 was chosen as this was the lowest value for which units could be readily identified using extraction techniques developed in this laboratory (8).

For each electrode, the total number of units identified and their signal-to-noise ratios were recorded and monitored as a function of time.

### *Histology*

After the final recording session, the cats were perfused with a buffered aldehyde solution. The fascicles with the implants were dissected free, osmicated, dehydrated, and embedded in plastic (Araldite). Semithin (0.5 $\mu$m) sections were cut at three levels: a) the implant site; b) 1.0–1.5 cm proximal; and c) 1.0–1.5 cm distal to the recording site. Control tissue was taken at equivalent levels from the radial nerves of cats that had no implanted electrodes.

Myelinated axons were counted and their diameters measured in each section. To determine whether a specifiable range existed over which electrodes could detect unit activity, proximity of the electrode to the nearest A$\alpha$ fibers was measured in each implant section.

## RESULTS

### *Electrode Stability*

Figure 3 shows the percentage of functional electrodes, fabricated as described in the Methods, as a function of time. At six months, six of the eight implanted bipolar electrodes remained functional. Failure of this set of electrodes was due solely to breakage of the leads, and not an inability to record action potentials with intact electrodes.

Figure 4 shows electrode impedance as a function of time for untreated electrodes in saline, and platinized electrodes implanted both inside and outside the fascicle. Unplatinized wires immersed in saline showed a steady increase in impedance over time.

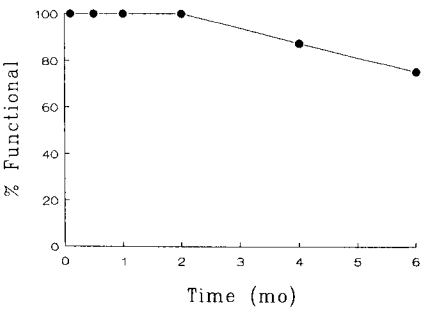


**FIGURE 3. Electrode survival rate. The percentage of functional electrodes is plotted versus time after implantation. This plot is based on data from eight electrodes constructed as described in the Methods section.**

Platinized nerve implants showed a small increase in impedance during the first month, after which the impedance remained stable.

## *Recording Properties*

Examples of action potentials recorded with these electrodes have been published elsewhere (8). The average number of active units per electrode is shown in Fig. 5. On the average, each electrode recorded activity with a signal-to-noise ratio greater than 1.4 from about 10 units in each recording session. Figure 6 shows the signal-to-noise ratio with respect to time. The lower curve shows the average of the mean signal-to-noise ratios for each functional electrode. A noticeable decrease in signal-to-noise ratio appears to occur after four months. Since the number of units was not the same for all electrodes, it was possible that the decrease might have been due to one or two

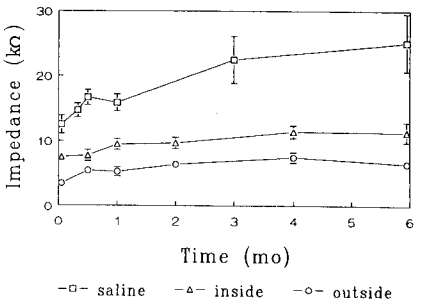


**FIGURE 4. Electrode impedance as a function of time. Impedances were measured with respect to ground. Plotted are the mean (± standard error of the mean) impedances for three untreated bipolar electrodes (squares) immersed in saline and twelve platinized electrode pairs implanted in the nerve (triangles for the intrafascicular wire, circles for the extrafascicular wire).**

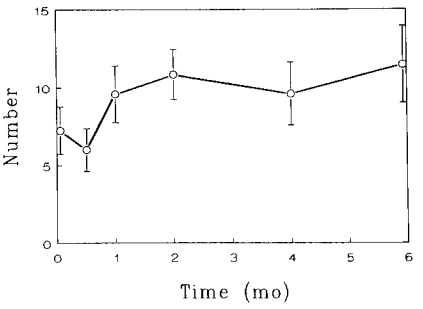


**FIGURE 5. Average number of units per electrode as a function of time. Only units with a signal-to-noise ratio greater than 1.4 have been included. Shown are means and standard errors.**

of the electrodes. To test for this, the mean signal-to-noise ratio of all identified units is shown in the upper curve. Both curves demonstrate small but statistically significant ($p < 0.01$) decreases in signal-to-noise ratio after four months. The noise level remained stable during the six-month period, averaging 5 to 6 $\mu$V. Thus, it appears the decrease in signal-to-noise ratio was due to a decrease in the size of the action potentials.

## *Histology*

Table 1 shows the average median diameter of myelinated axons for nine normal control radial nerves. Table 1 also shows the corresponding data set for 11 of the implants (adequate sections were not obtained from one of the implanted fascicles). Comparison of the three levels of the implanted nerves with the control nerves demonstrated a 40% ($p < 0.001$) reduction in axon size at the level of the implant, primarily due to a reduction in the degree of myelination in proximity to the implant, but essentially normal values for axons proximal and distal to the implant zone.

In eight implants the distance from the electrode to the nearest A$\alpha$ fibers was mea-

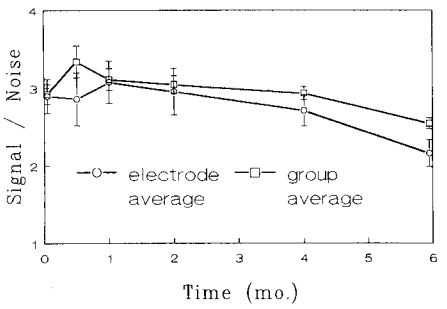


**FIGURE 6. Signal-to-noise ratio over time. The lower curve (electrode average) represents the average of the mean signal-to-noise ratios for the functional electrodes. The upper curve (group average) shows the signal-to-noise ratio averaged over all units. Shown are means and standard errors.**

**TABLE 1. Axon diameters for control and implanted tissue from sections taken proximal to, through, and distal to the implant zone. The averages, with standard deviations for the median axonal diameters, as well as the diameters below which 5 and 95% of the population fell are given, normalized to the mean of the proximal median diameter of the control nerves.**

| | Control | Implant |
|---|---|---|
| Proximal: | | |
| 5% | 0.3 ± 0.0 | 0.3 ± 0.1 |
| Median | 1.0 ± 0.2 | 0.9 ± 0.1 |
| 95% | 1.9 ± 0.2 | 1.7 ± 0.1 |
| Implant zone: | | |
| 5% | | 0.2 ± 0.1 |
| Median | | 0.6 ± 0.1 |
| 95% | | 1.4 ± 0.3 |
| Distal: | | |
| 5% | 0.3 ± 0.1 | 0.3 ± 0.1 |
| Median | 1.1 ± 0.2 | 0.8 ± 0.1 |
| 95% | 1.9 ± 0.2 | 1.5 ± 0.2 |

sured. At least 10 A$\alpha$ fibers occurred within 65 $\mu$m from the electrode surface of all implants, even for those in which there were fewer than 10 recordable units at 6 months. This suggests that differences in number of units and signal-to-noise ratio between implants may be due to the electrical properties of connective tissue around the electrode rather than tissue damage caused by the electrode.

Minor reactive changes occurred around the six-month implants that were histologically examined. Representative examples of the recording site and the Teflon insulated portions of the electrodes are illustrated in Fig. 7. A mild foreign body reaction developed at the platinum and Teflon surfaces which is characteristic of biocompatible materials (17,19,20). It consists of a relatively thin layer of multinucleate giant cells lacking the cytoplasmic inclusions indicative of active phagocytosis. Connective tissue is the second component of the capsule which consists of alternating layers of collagen and fibroblasts in a concentric array whose thickness is indicative of the size of the implant. The fact that this layer is less than 50 $\mu$m thick is consistent with the small size and modest movement of the implant. The absence of granular leukocytes is a further indication of the lack of an inflammatory reaction in the capsule and the general stability of the implant within the fascicle.

Relatively minor reactive changes are also evident in endoneurium surrounding the electrodes. As shown in Fig. 7, there is an increase in endoneurial connective tissue near the surface of the capsule and a reduction in the incidence of myelinated axons. In some instances this is also associated with a slight reduction in the size of axons, and with thinning of the myelin sheath (Fig. 7A). In some nerves, a few regenerating units occurred within the fascicle bearing the electrode, indicating surgical trauma at the time of the implantation (Fig. 7B).

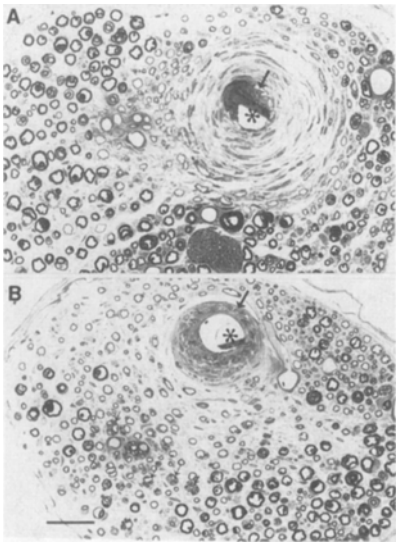


**FIGURE 7. (a) Semithin section of a radial nerve fascicle illustrating the interface with an implanted electrode at the uninsulated recording site. The platinum wire (asterisk) of this six-month implant is folded over and is surrounded by both a dark multinucleate giant cell and by a thick connective tissue capsule. Relatively minor changes occur within the endoneurium consisting of a reduction in the incidence of myelinated axons, and thinning of their myelin sheath. (b) Semithin section of a radial nerve surrounding the insulated portion of a six-month electrode implant (asterisk). Although the capsule is thin, and there is some reduction in the incidence and size of myelinated axons, there are also a few regenerating units (arrows) indicating slight injury and subsequent regeneration of a few axons. The calibration bar indicates 50$\mu$m for both plates.**

## DISCUSSION

### *Intrafascicular Electrodes as Sources of Sensory Information*

The results of this study and its companion (8) show that the activity of individual, functionally identifiable nerve fibers can be recorded for up to six months with these intrafascicular electrodes. Sensory fibers from muscle spindles and tendon organs are similar in size to the cutaneous fibers recorded from in the present study (13), so our results can be extended to muscle sensory receptors as well. We have also shown that such electrodes provide a representative sample of the different types of functional elements within a given fascicle, providing potentially useful information for the control of functional neuromuscular stimulation systems (8).

Of particular concern with this method, however, is the optimum number of units to be sampled by a single electrode. Peripheral mechanoreceptors innervated by A$\alpha$ fibers will fire at rates on the order of 100 impulses/s when activated by moderately strong stimuli, and the duration of individual action potentials seen with intrafascicular electrodes are typically on the order of 1 ms. We suggest, therefore, that the number of units from which activity can be recorded by a single electrode should not significantly exceed 10 to avoid significant superposition and to preserve the identity of individual units. The intrafascicular electrodes employed for this study record from

about 10 units, a number small enough to allow unit separation and identification yet large enough to provide a reasonable sample of fascicular activity.

In distal portions of the extremity, the organization of fascicles is such that one can obtain information from specific muscles or restricted areas of the skin. In proximal portions of a nerve, it should be possible to implant multiple electrodes and to monitor activity in several populations of axons without compromising the integrity of the nerve.

Intrafascicular electrodes may also provide a method whereby sensation could be restored to paralyzed patients, if a means can be found to deliver the information to somatosensory cortex. Such restoration of sensation is a vital, but little considered, component of functional restoration. Clinical experience has shown that individuals with full-motor control, but no sensation in a limb, will not use the limb if the other limb is sensate (4,21). This implies that even if FES can restore lost function, the patient may resist using a limb which is not felt. Cutaneous somatosensory afferent activity recorded with our electrodes could be processed and used to stimulate the sensory cortex via an intracortical electrode array. Such an array is under development here for an artificial vision program (1).

### *Electrode Characteristics*

The signal-to-noise ratio of electrodes experienced a slow, but significant decrease over the six-month period of the study. Although the signal-to-noise ratio remained high enough to allow ready identification of single units and the level of the background noise remained relatively constant, extrapolation of the data projects an expected functional lifetime for these implants of some 18 to 24 months if this trend were to continue. On the other hand, if no further electrode encapsulation were to occur, the electrodes might function indefinitely.

If implantation of the electrodes, or their presence alone, was significantly injurious to neural tissue, the recordings should have failed early and completely. However, this was not the case. Furthermore, the histology of the implants indicated that the electrode site is well encapsulated and that nearby neural tissue remains viable. On the other hand, shifts in the recorded population suggest continual, small dislocations in the position of the electrode over time (8). Tissue responses to these movements could explain the observed slow decline in signal-to-noise ratio. Elimination of these movements, perhaps by fabricating the electrodes with more flexible materials or fixing the nerve with respect to the wire, should further improve the longevity of these electrodes.

In general, the reactive changes occurring after six months of implantation were similar to those around small biocompatible implants in other parts of the body (3,17,19,20). They are consistent with the notion that a stable relationship is established soon after implantation between the electrode and nearby axons and that there is no appreciable inflammatory reaction which would compromise the physiological status of nerve conduction in proximity to the electrodes.

### *Comparison with Other Recording Electrodes*

Two other devices have been proposed for chronic recording of neural activity from peripheral nerves. Regeneration electrodes consist of microelectrode arrays

through which transected nerve fibers are expected to grow (5,7). Theoretically, such electrodes are extremely selective, permitting individual recording from large numbers of nerve fibers, and no spike separation process is required. However, regeneration electrodes require transection of a healthy nerve and subsequent successful regeneration. Given the limited ability of transected nerve fibers to restore functional connections with the periphery, it is doubtful that enough fibers would successfully regenerate through such an array to justify intentionally compromising an intact nerve.

Cuff electrodes, on the other hand, have an established history of providing long-term recording of neural activity (9,18). A major advantage of cuff electrodes is their ability to eliminate electromyogenic (EMG) noise. However, they record activity from an entire nerve as a weighted average, rather than as identifiable units. Only in special cases can the sources contributing to such a composite signal be identified. Another drawback is that cuff electrodes must be properly fitted for each nerve, otherwise the blood supply to the nerve may be compromised by localized swelling within the relatively stiff polymer cuff surrounding the nerve. A recently developed spiral cuff electrode (15) may offer a solution to this latter problem, but it does not address the problem of lack of specificity when recording whole nerve activity.

The intrafascicular electrode described in this study does not require transection of the nerve, causes only modest damage during implantation, and does not compromise the blood supply to the nerve. Its bipolar design should limit interference from EMG signals when the electrodes are implanted in proximal nerve fascicles, although this has not been explicitly tested. Our intrafascicular electrode can be implanted in virtually any fascicle without requiring a custom fit for fascicles of different diameters and can provide adequate selectivity for single-unit identification, in contrast to cuff electrodes which, by their nature, make it nearly impossible to detect single-unit activity.

## SUMMARY

Based on the evidence presented in this report we believe that further development of intrafascicular electrodes is warranted. Light micrographs indicate a high degree of biocompatibility and suggest that the modest reactive changes are due to differential movement of the relatively stiff electrode within the fascicle. Production of an electrode whose physical properties are more similar to components of the nerve, by reducing this movement, should yield an intrafascicular electrode that is more stable and long lived.

## REFERENCES

1. Campbell, P.K.; Normann, R.A.; Horch, K.W.; Stensaas, S.S. A chronic intracortical electrode array: Preliminary results. J. Biomed. Mater. Res.: Applied Biomaterials 23:245–259; 1989.
2. Crago, P.E.; Chizeck, H.J.; Neuman, M.R.; Hambrecht, F.T. Sensors for use with functional neuromuscular stimulation. IEEE Trans. on Biomed. Eng. 33:256–268; 1986.
3. Delgado, J.M.R. Chronic implantation of intracerebral electrodes in animals. In: Sheer, B.E., ed. Electrical stimulation of brain. Austin, TX: University of Texas Press; 1961: pp. 25–36.
4. Dellon, A.L. Evaluation of sensibility and reeducation of sensation in the hand. Baltimore, MD: Williams and Wilkins; 1981.
5. Edell, D.J. A peripheral nerve information transducer for amputees: Long-term multichannel recordings from rabbit peripheral nerves. IEEE Trans. on Biomed. Eng. 33:203–214; 1986.

6. Edell, D.J.; Wolfe, J.H. Sensing of neural potentials using micromachined electrode arrays. IEEE/ONR/NIMH Workshop on Synthetic Microstructures in Biology and Medicine. Mar. 24–26, 1986.
7. Edell, D.J.; Churchill, J.N.; Gourley, I.M. Biocompatibility of a silicon based peripheral nerve electrode. Biomater. Med. Devices, Artif. Organs 10:103–122; 1982.
8. Goodall, E.V.; Horch, K.W.; Lefurge, T.M. Information contained in sensory nerve recordings made with intrafascicular electrodes. IEEE Trans. on Biomed. Eng. (in press).
9. Hoffer, J.A.; Loeb, G.E.; Pratt, C.A. Single unit conduction velocities from averaged nerve cuff electrode records in freely moving cats. J. Neurosci. Methods 4:211–225; 1981.
10. Hoffer, J.A.; Loeb, G.E. Implantable electrical and mechanical interfaces with nerve and muscle. Ann. Biomed. Eng. 8:351–360; 1980.
11. Horch, K.W.; Tuckett, R.P.; Burgess, P.R. A key to the classification of cutaneous mechanoreceptors. J. Invest. Derm. 69:75–82; 1977.
12. Malagodi, M.S.; Horch, K.W.; Schoenberg, A.A. An intrafascicular electrode for recording of action potentials in peripheral nerves. Ann. Biomed. Eng. 17:397–410; 1989.
13. Mann, M.D. The nervous system and behavior. Philadelphia, PA: Harper & Row; 1981.
14. Nannini, N.; Horch, K.W. Muscle recruitment with intrafascicular electrodes. IEEE Trans. on Biomed. Eng. (in press).
15. Naples, G.G.; Mortimer, J.T.; Scheiner, A.; Sweeny, J.D. A spiral nerve cuff electrode for peripheral nerve stimulation. IEEE Trans. on Biomed. Eng. 35:905–915; 1988.
16. Peckham, P.H. Functional electrical stimulation: Current status and future prospects of applications to the neuromuscular system in spinal cord injury. Paraplegia 25:279–288; 1987.
17. Schultz, R.L.; Willey, T.J. The ultrastructure of the sheath around chronically implanted electrodes in brain. J. Neurocytol. 5:621–642; 1976.
18. Stein, R.B.; Nichols, T.R.; Jhamandas, J.; Davis, L.; Charles, D. Stable long-term recordings from cat peripheral nerves. Brain Res. 128:21–38; 1977.
19. Stensaas, S.S.; Stensaas, L.J. The reaction of the cerebral cortex to chronically implanted plastic needles. ACTA Neuropath. (Berl.) 35:187–203; 1976.
20. Stensaas, S.S.; Stensaas, L.J. Histopathological evaluation of materials implanted in the cerebral cortex. ACTA Neuropath. (Berl.) 41:145–155; 1978.
21. Sunderland, S. Nerves and nerve injuries. Baltimore, MD: Williams and Wilkins; 1968.