BASIC DESIGN CONSIDERATIONS FOR CHRONICALLY IMPLANTABLE NEURAL INFORMATION SENSORS

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Chronically implantable neural information sensors are designed to transduce electrical potentials from the central or peripheral nervous system. The purpose of developing this technology is to eventually provide information to control assistive devices for the physically disabled. The technology will potentially benefit a variety of disabled individuals such as amputees and spinal cord injury patients.

Current research focusses on the design and fabrication of the structure used to interface electrical systems with nerve tissue. The following discussion highlights some of the important design considerations involved in the development of this technology.

Neural activity consists of 250 microsecond current pulses (action currents) which flow through active cell membranes into extracellular spaces surrounding cell bodies and axons in the nervous system. Current density at the active membrane surface is approximately .01A/cm^2 [1]. The action currents create potential gradients (action potentials) which can be sensed by electrode contacts in the extracellular space. The electrode arrays in figures 1 and 2 can be used to sense such potentials.

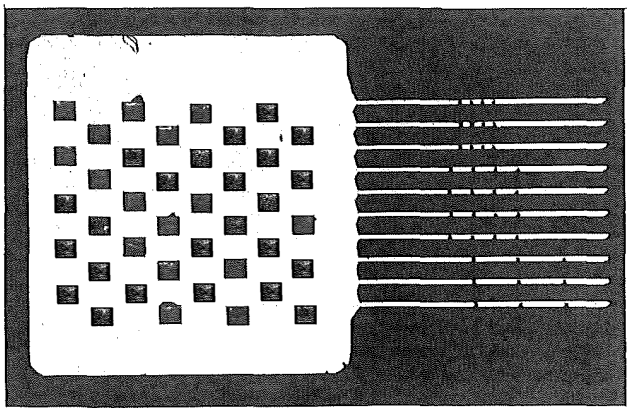

Figure 1: Insertable 40 contact electrode array.

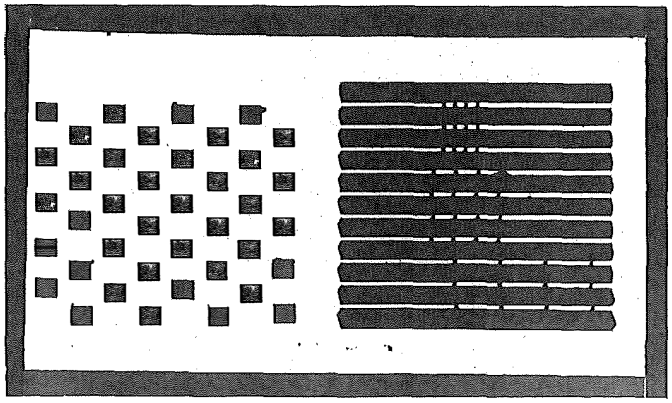

Figure 2: Regeneration 40 contact electrode array.

These arrays were chemically machined from (110) silicon substrates. The silicon bars are 40 microns wide and 40 microns thick. Each bar is 2mm long and carries four electrode contacts spaced from 100 to 400 microns apart. The bars are spaced 200 microns apart. Four micron wide, phosphorus doped polysilicon conductors connect the bonding pads with the electrode contacts. Polysilicon conductors were necessary to reduce thermal stress which can induce crystal defects in the etched silicon bars near the end of the conductors.

Implantable electrode arrays must be biocompatible and must be able to withstand immersion in corrosive physiological environments indefinitely. The metal system used on these arrays was selected after evaluating a number of potential systems in heated, oxygenated Ringer's solution [2]. Tantalum was used to form a stable, ohmic, Ta-TaSi-Si contact. A platinum layer formed a stable, oxidation resistant interface to the tantalum. The outer metal was gold which was the most biocompatible metal available. The gold was deposited with sufficient surface energy to enhance nucleation and produce a micro-rough surface. The micro-rough surface reduced the magnitude of the electrode/electrolyte impedance to 340MΩ-μ^2 at 1Khz. A combination of LPCVD silicon nitride and APCVD silicon dioxide was deposited to insulate the conductors.

The biocompatibility of the structures has been verified in the peripheral nervous system and the cerebral cortex of rabbits [3,4]. Biocompatiblity is a relative term. Certainly, devices which produce gross inflammatory response are not biocompatible. Devices which may not cause obvious inflammatory response but which cause neuron cell death are not biocompatible. The definition is less obvious, however, when the degree of connective tissue encapsulation of the implant is considered. All physical structures implanted in mammalian systems are encapsulated by varying thicknesses of connective tissue. This connective tissue displaces the cells being sensed from the electrode contacts which reduces signal amplitude and electrode selectivity. Thus the degree of connective tissue encapsulation is of critical importance because it influences design of the electrode contacts themselves.

Cortical Electrodes

Electrode contact area is the primary design variable which can be optimized for signal, noise and selectivity. In the cerebral cortex, the activity of neuron cell bodies is of interest. There are approximately 24,000 cell bodies per mm^3 in the cortex. The cortex is 1.5 to 4.5 mm thick. Cell diameters range from 6 to 70 microns in the cortex [5]. Simple analyses assume that these cells represent spherical current sources in a homogeneously conductive medium. The maxmimum potential seen at the surface of a cell membrane relative to a distant reference is given by:

$$V_{max} = pJr_s$$

where: p = effective resistivity of the medium
J = membrane current density
r_s = radius of the cell

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According to this simple analysis, a 1.5mV potential should be seen at the surface of a 15 micron radius cell in a 100Ω-cm medium assuming a membrane current density of .01A/cm^2.

Under the above assumptions, potentials from action currents can be shown to vary as 1/distance from the center of the source. The potential seen by a point contact in the medium is given by:

$$V = \frac{\rho J r_s^2}{r}$$

where: r = radial distance from effective center of the source.

The actual spacing of the neuron cell bodies from the electrode contacts is determined by the electrode orientation and the amount of encapsulation by glial cells. Recent biocompatibility tests of the insertable electrode array have shown that a 5 micron microglial cell layer typically encapsulates the array.

A point contact on the electrode array located 5 microns from a 30 micron diameter cell would be expected to sense a 1.1mV potential during the peak of an action potential. A 6 micron cell for the same conditions would generate a potential of 110μV.

The electrode contact area does not significantly influence signal amplitude as long as electrode contacts are relatively concentric with the equipotentials. However, noise and selectivity are inverse functions of electrode size. The choice of appropriate electrode size then depends on acceptable noise versus acceptable selectivity. The larger the electrode area, the more likely it will overlap significant potentials from adjacent cells. Exactly how large an electrode can be depends on required signal differentials for the instrumentation, cell sizes and spacing, and connective tissue thickness.

Maximum selectivity can be achieved by minimizing the geometric electrode area until electrode impedance becomes the limiting factor. Since minimization of electrode area maximizes the number of electrodes possible on a given structure, setting the electrode area on the basis of electrode impedance levels is a reasonable approach. (It should be noted here that the degree of selectivity appropriate for neural prostheses is not well defined at present. Less selective electrodes have the advantage of integrating activity from several related sources which may improve efficiency of data acquisition considerably). At present, it appears that the peak to peak noise level of the input electrodes should be less than 20 microvolts to avoid signal degradation. A plot of noise versus electrode radius for the rough gold system is shown in figure 3.

From this analysis, a minimum electrode area of 150 μ^2 would be required to satisfy noise requirements. An electrode which was 12x12 microns should easily record from 6u diameter cells spaced 5μ from the electrode plane (effective diameter = 16μ). Such an electrode would be expected to generally sense activity from one or two closely spaced 6u cells, but up to three cells could conceivably create significant potentials on the electrode. It is likely that such selectivity is adequate for most practical purposes at present.

An electrode area of 150μ^2 would be expected to have an impedance of 2.3 Megohms at 1 kHz. The presence of such a high source impedance is the primary driving force for integrating circuitry onto electrode structures used for chronic implantation.

Peripheral Nerve Electrodes

In the peripheral nervous system, activity from myelinated axons is of interest for control of prostheses for amputees. There are roughly 10^4 axons/mm^2 arranged longitudinally in bundles to form peripheral nerves. Activity from these axons can be sensed near active membrane nodes. Nodes are 1 micron wide swaths of electrogenic membrane which encircle the nerve at longitudinal spacings of 300 to 3000 microns.

The regeneration electrode structure in figure 2 can be used to sense activity from these nodes [2]. Transected peripheral nerves grow through the slots in the implant. The electrode contacts lie transversely oriented to the axons. This results in a weak relationship between electrode radius and recorded potentials. Because of the minimum of 10 microns of connective tissue encapsulation observed during biocompatibility testing, nodes cannot be closer than 10 microns to any contact. Since internodal spacing is 300-500 microns in regenerated rabbit peripheral nerve, the probability that a particular axon node will be located within ten microns of the plane of the electrode array is about .025. Relatively few axons would be expected to yield maximal signals on a given electrode contact with this design. Electrodes tend to be selective, then, because of this random spacing of nodes relative to the electrode plane.

In spite of uncertainties in absolute values, some insight into the design of electrode contacts for the regeneration arrays for peripheral nerves can be gained by performing an analysis with an idealized model of the system [2]. The results of such an analysis (figure 3) point out that noise may limit the usefulness of an electrode configuration, and that there is a minimum size requirement for electrode contacts. This is because signal amplitudes are a weak function of electrode radius while peak electrode thermal noise is inversely proportional to electrode radius.

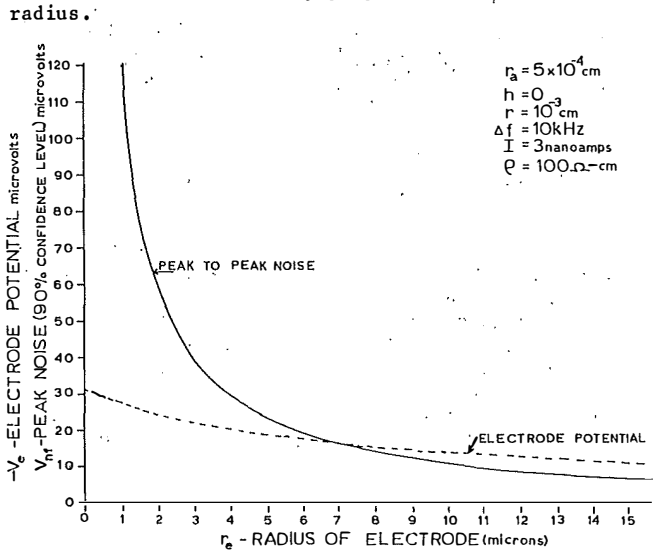

Figure 3: Electrode potential and peak noise as a function of electrode radius for rough gold metallization.

Experimentally, the maximum recorded potentials (figure 4) are an order of magnitude larger than predicted by a model which assumes a homogeneous medium.

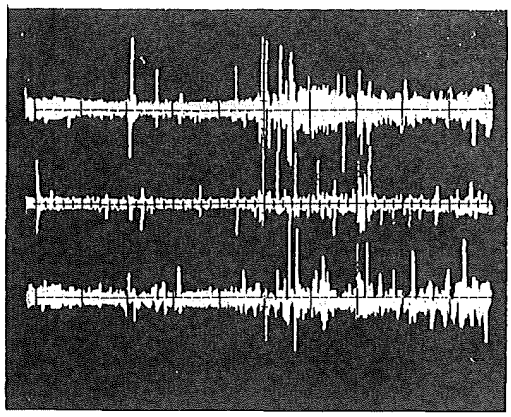

Figure 4: Three sets of differentially recorded peripheral nerve activity from a chronic silicon implant. Scale - Vertical = 500μV, Horizontal = 200msec

There are several possible explanations for the discrepancy. Histological analysis revealed that near the implant, fibroblasts formed hollow tubes which encircled small groups of axons (figure 5).

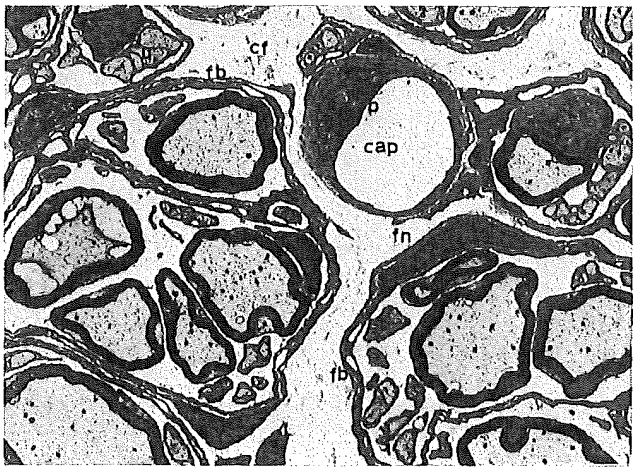

Figure 5: TEM of cross section of rabbit peripheral nerve which regenerated through a silicon grid one year earlier (2850x).

If these sheaths constrict the flow of action currents, then very large potentials may be observed within the tubes. Electrodes which do not intersect the sheaths would be expected to yield smaller signals than predicted. Electrodes which are connected to the sheaths may exhibit much larger potentials depending on the electrical properties of the elements and the structure of the connection. At this time, the electrical nature of these connective tissue sheaths is unknown.

References

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