Background on Silicon Substrate Probes

Our work on microelectrode arrays for single-unit recording in the central nervous system really began many years ago with early work done at Stanford [7,8,9]; however, more recent efforts began in the early 80s. From the beginning, this work was aimed at utilizing the broad base of process technology developed for silicon integrated circuits as closely as possible. This was done to ensure the use of processes that are high-yield and capable of high volume, to allow the eventual use of on-board signal processing circuits, and to utilize the capabilities for small size associated with such processes. The basic probe structure is shown in Figure 1. A micromachined silicon substrate supports an array of thin-film conductors that are insulated above and below by deposited dielectrics of silicon dioxide and silicon nitride. Openings in the upper dielectrics along the probe shank define stimulating or recording sites which are inlayed with gold or iridium oxide for interfacing to the tissue. At the rear of the probe, integrated circuitry provides amplification, filtering, and a multiplexed interface to the output leads, which connect to a remote percutaneous plug.

Virtually all of the efforts directed at photoengraved microelectrode arrays during the past quarter century have had the structure in Figure 1 in common; however, the various efforts have differed considerably in their choices of substrate materials, insulating dielectrics, and substrate shaping technologies. These choices have resulted in widely differing process yields and have sometimes compromised compatibility with batch processing. Most approaches have required double-sided processing of thin substrate materials (silicon, molybdenum, or polyimide), which has prevented the processes from being truly batch and has limited the dimensions associated with the final probes. Some efforts have used photore sist or polyimide as dielectrics, and these, generally, have not met with long-term (chronic) success. Only the silicon approaches have attempted to include electronics on-board, and it has only been recently that single-unit recording has been demonstrated. In addition, until recently no satisfactory output lead structure had been identified, preventing the realization of practical chronic assemblies. Thus, in approaching structures such as that shown in Figure 1, the critical areas to be considered are the substrate material (and how to shape it), the insulating dielectrics and their compatibility with the interconnect materials used, the possibility for merging the structure with electronics to reduce the number of output leads, and finally, the structure and encapsulation of the output leads themselves.

The most important single result from our work at Michigan has been the use of a diffused boron etch-stop for silicon substrate definition. This has allowed precise, reproducible substrate formation with a rounded probe cross-section, dimensional control to better than 1µm and a minimum shank width of 20µm or less. This is accomplished with a single-sided process using wafers of standard thickness. No other process we have considered has allowed this degree of control in a high-yield batch process. Secondly, we have shown that oxide/nitride stress-compensated dielectrics, in combination with interconnects of polysilicon, tantalum, or refractory silicides, can provide reliable encapsulation for the high-impedance recording sites for durations of at least many months and probably much longer. Site impedances, for example, remain stable over many months in-vivo in the range of a few megohms. Thirdly, we have demonstrated the incorporation of high-performance CMOS circuitry in the probe substrates, both for recording and for stimulation. This is the first time such circuitry has been integrated to form active probes capable of recording from single units in-vivo. Finally, we have demonstrated the use of the basic probe process to form multistrand multiconductor silicon ribbon cables for use in connecting the probes to the outside world. These cables offer minimal tethering forces, can be built directly into the probe itself, and can be completely surrounded by a conducting shield, protecting the dielectrics from the extracellular fluid. These cables have allowed chronic probes to be fabricated and used in-vivo for periods exceeding six months.

Basic Passive Probe Structure and Fabrication

Passive electrode arrays do not include any on-chip circuitry. Input pads on the rear portion of the probe substrate make contact to thin-film conductors that lead down the probe shank(s) to the recording sites spaced along them. More than 20 different designs can be accommodated on a typical mask,
depending on the sizes of the probes needed. Shank widths as narrow as 15µm and length-to-width aspect ratios of more than 100:1 have been realized using the basic probe process.

The sequence of steps making up the overall fabrication process for these passive probes is shown in Figure 2. Fabrication begins with a standard silicon wafer that is oxidized and patterned to define the intended probe areas. These areas are subjected to a deep boron diffusion (15 hours at 1175°C) to heavily dope the probe substrate. The diffusion time and temperature are selected to produce a final probe thickness of about 15µm. The masking oxide is then stripped and the lower dielectric layers are deposited using low-pressure chemical vapor deposition (LPCVD). These films consist of a layer of silicon nitride sandwiched between two layers of silicon dioxide. The thickness of each layer is carefully selected so that the thermal expansion coefficient of the composite insulator matches that of silicon to avoid warp in the final probes.

Conductors of phosphorus-doped polysilicon are next deposited and patterned, followed by the deposition of upper dielectrics which are identical in composition and thickness to the lower films. The upper dielectrics are now patterned using a dry etching process (reactive ion etching (RIE)) to open passivation sites for the recording/stimulation sites and bonding pads. Using liftoff techniques, gold is inlayed to form the bond pads and iridium to form the recording/stimulation sites.

The field dielectrics outside of the intended probe areas are now removed using RIE. Finally, the wafer is thinned to about 100µm from the back in an isotropic etchant (10% hydrofluoric, 90% nitric acid) and is then subjected to an unmasked etch in ethylene diamine-pyrocatechol-water (EDP) [2] to separate the individual probes. The EDP etches silicon, stopping on the heavily-doped boron regions. It does not attack the dielectrics or metals used in this process. The completed probe chips are removed from the etch, ready for the lead attachment and mounting. An etched out stimulation probe is shown on the head of a penny in Figure 3.

This simple eight-mask process is capable of yields in excess of 80%, results in very small structures, and requires only single-sided processing on standard silicon wafers. All etching steps are highly selective and self-stopping. Probe features can be controlled to within 1µm or better. The substrate can have any two-dimensional shape as defined by the mask. All probe materials are biocompatible and suitable for chronic applications [1,3,4].

Silicon Ribbon Cables

In a chronic system, the interconnect performs the critical link between the microelectrode to the external world through a percutaneous connector. We have developed miniature, flexible, multilead silicon ribbon cables fabricated using the same photolithographic techniques and basic process as for the probes themselves.

Figure 4 shows the structure of the ribbon cable. It consists of a long, thin silicon substrate that supports multiple dielectrically-encapsulated leads. At either end of the cable is a thicker platform with exposed metal pads for bonding the cable to either a probe or to a percutaneous connector. The main cable itself can be further passivated with an outer barrier layer (typically gold or polysilicon) over the upper dielectrics. This layer makes contact to the silicon substrate so that the leads are electrically as well as chemically shielded, making the cable in effect a multilead "coaxial" structure. The ribbon cable process is completely compatible with the passive probe process. It only requires one extra mask step to define the cable region, which is formed using a shallow boron diffusion. In fact, one of the greatest advantages of silicon ribbon cables is that they can be integrated into the probe itself, eliminating the need for any bonding, soldering, or encapsulation between the probe and the interconnect.

The final thickness of the cable is 4-5µm. These cables have excellent flexibility in all directions. They can routinely be tied in "knots" with radii of less than 400µm. Their ability to twist in the plane of the cable can be further improved by slotting the cable into multiple strands (by masking the boron diffusion appropriately), each supporting individual conductors. A chronic assembly made up of an integrated cable/probe and a percutaneous connector (Microtech, Inc., FR-12S-4) is shown in Figure 5.

The electrical integrity of the cables has been characterized in vitro by soak tests in buffered saline. Soak tests under electrical stresses of +5V and -5V have shown that the cables exhibit sub-picoamp leakage currents for over 4 years, with no measurable degradation in any characteristic. Long-term in vivo experiments in guinea pig inferior colliculus have shown that chronic assemblies using the cables can remain functional for periods of at least one year. While still longer-term experiments are needed, these cables have opened the way to a great many physiological investigations which were not otherwise possible with the probes.

References

Implant assembly currently used for chronic guinea pig preparations. Shown here is the underside of a percutaneous connector joined to a five-site probe by a single-strand ribbon cable.

Figure 4 [click to enlarge]

Figure 5 [click to enlarge]


