A Micromachined Silicon Depth Probe for Multichannel Neural Recording

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***Abstract*—A process of making a new type of silicon depth-probe microelectrode array is described using a combination of plasma and wet etch. The plasma etch, which is done using a low temperature oxide (LTO) mask, enables probe thickness to be controlled over a range from 5 to 90 μ. Bending tests show that the probe's mechanical strength depends largely on shank thickness. More force can be applied to thicker shanks while thinner shanks are more flexible. One can then choose a thickness and corresponding mechanical strength using the process developed. The entire probe shaping process is performed only at low temperature, and thus is consistent with the standard CMOS fabrication. Using the probe in recording from rat's somatosensory cortex, we obtained four channel simultaneous recordings which showed clear independence among channels with a signal-to-noise ratio performance comparable with that obtained using other devices.**

***Index Terms*—Mechanical stress test, multichannel simultaneous recording, neural probes, plasma etch, silicon microelectrode array.**

I. Introduction

THE use of the microelectrode to record the electrical activity of single neurons has contributed much to our understanding of the nervous system. Simultaneous recording from populations of neurons has been increasingly acknowledged as essential to better understanding of functional relationships between cells in brain and other neuronal networks [1]. For this purpose, thin film microelectrode arrays have many advantages over wire bundles. 1) Using thin film processing, we can arrange multiple sensing sites along a single shaft of silicon, reducing device volume and tissue damage as compared to multiple electrodes. 2) Advanced photolithography permits a good control of electrode dimensions and relative positions with sub-micron precision, while guarantees uniform and repeatable electrode characteristics. 3) We can integrate the electrode with on chip circuitry because we use a silicon substrate [2].

Seeing these clear advantages, a number of research groups have been developing silicon-based microelectrode arrays (microprobes) since the late 1960's [3]–[11]. Some used anisotropic wet etching for probe shaping [3], [4]. But this method does not give reproducible and uniform probe dimensions. Others have employed high temperature boron diffusion and selective wet etch to define the precise probe shape [5]–[7]. Recently, the combination of front side plasma etch and backside wet etch was also introduced for probe shaping [10]. Considering the wide availability of the CMOS foundry services such as MOSIS for custom integrated circuits, one may prefer the latter approach that involves neither high temperature nor impurity doping. Probe shaping can then be performed as a post process to the CMOS fabrication. This method also allows wider control over the probe thickness than the boron doping technique.

The depth probe (called "SNU probe" hereafter) described in this paper was also fabricated using combination of dry (plasma) and wet etch. Using plasma etch with readily available low temperature oxide (LTO) as an etch mask, we were able to obtain a 30-μm probe thickness, well beyond that readily obtainable by boron diffusion. Using the LTO mask, we could avoid problems associated with thick photoresist (PR) which was used previously [10]. The mechanical and electrical characteristics of the SNU probe have been studied. SNU probes with 30-μm-thick shank were used to record neural activity from rat's somatosensory cortex. Under the same experimental conditions, conventional tungsten electrodes and probes made by Univ. of Michigan (called "Michigan probes" hereafter) were also tested for comparison.

II. Methods

A. Probe Fabrication

A thin tapered silicon shaft (shank) supports an array of metal recording sites and interconnects that reach bonding pads situated on carrier area. In the carrier area, the original wafer thickness remains, which facilitates subsequent handling of the probes. The metal layer is passivated upon and below by an oxide/nitride/oxide triple dielectric layer, which acts as an effective barrier of ions and water in saline environment [11].

Fig. 1 shows the overall process flow developed. The starting material was $\langle 100 \rangle$ oriented silicon wafers (p-type, 530-μm thick) polished on both front and back sides. After 100 nm of thermal silicon dioxide and 200 nm of LPCVD silicon nitride were deposited on both front and back sides, 800 nm of PECVD

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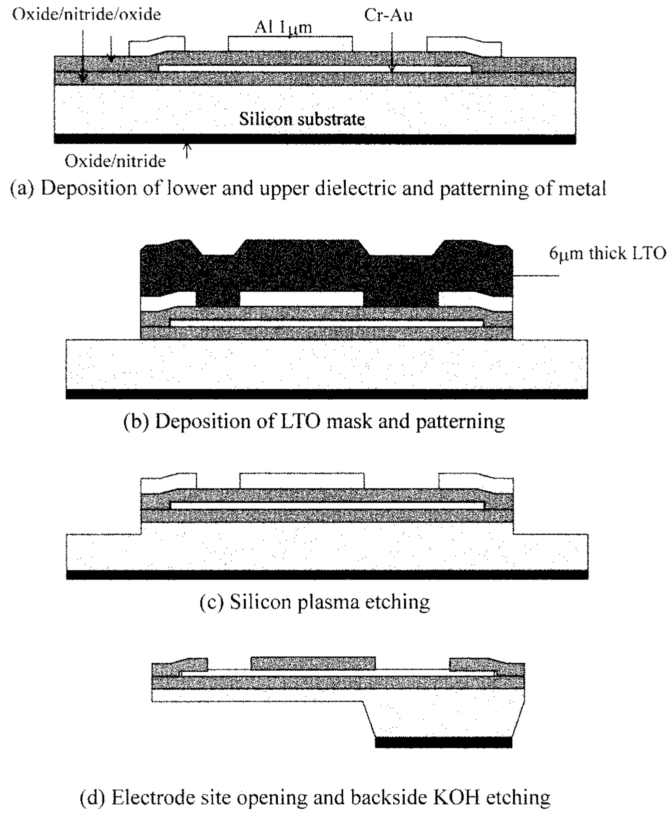


Fig. 1. Overall process flow of the depth probe developed using LTO/dry etch process.

silicon dioxide was deposited on the front side to form the triple lower dielectric. Oxide/nitride layers on back side were later patterned to form the masking layer for the KOH etch process.

The gold metallization layer was then patterned for recording sites, bonding pads and interconnections. We used image reversal of the PR layer (AZ5214) to obtain an overcut structure which is suitable for the lift-off process. Over the patterned PR structure, 100 nm of chrome and 300 nm of gold were evaporated sequentially. The resist was then dissolved away in acetone, lifting off the unwanted metal with it.

The second triple dielectric layer consisting of PECVD silicon dioxide (200 nm), silicon nitride (200 nm), and silicon dioxide (800 nm) was then deposited. A 1-μm-thick aluminum masking layer was patterned for opening the windows of recording sites and bonding pads. This aluminum masking layer was patterned at this point to avoid the photo process problems associated with non planar structures if the patterning were to be done after deep plasma etch.

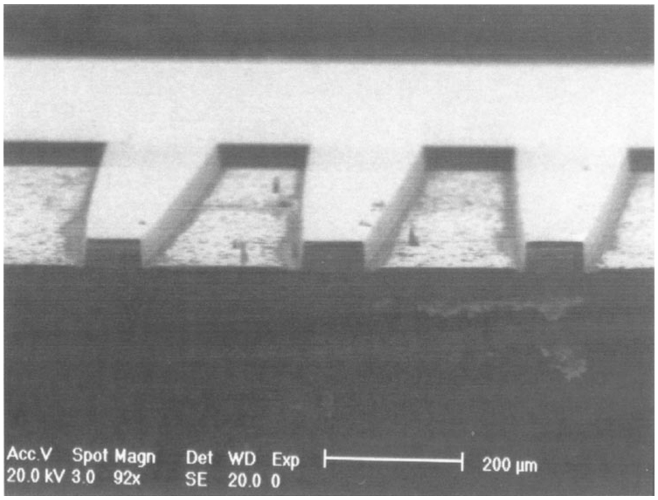
A 6-μm-thick PECVD LTO masking layer for the deep plasma etch process was then deposited and patterned. The etch rate of LTO using CHF_3 and CF_4 gases was 60 Å/s for a chamber pressure of 130 mtorr at the RF power of 600 W under the magnetic field of 60 gauss. Subsequently the deep etch was done using Cl_2 reactant gas to a depth of 30 μm. A Cl_2 gas flow of 40 sccm was used with a chamber pressure of 7 mtorr at the RF power of 300 W. The etch rate of Si using Cl_2 gas was 4250 Å/min. The latter etch depth determines the final shank thickness. The process done on the front side was completed by the opening of windows for electrode sites and bond pads using plasma etch, and the removal of the Al masking layer.

The back side oxide/nitride masking layer was patterned using double side alignment. The wafer was then placed in a 30 wt% KOH solution at 70°C with the front side protected using black wax (Apiezon W). The black wax was removed using Trichloroethylene (TCE). After the final probe shaping, the probes were mounted and wire-bonded to a printed circuit board (PCB) which was designed for easy connection to an external electronic circuit system. Terminal connection on the PCB was made in a standard DIP socket. Coating the wire bonded area with Sylgard (Dow Corning) completed the fabrication process.

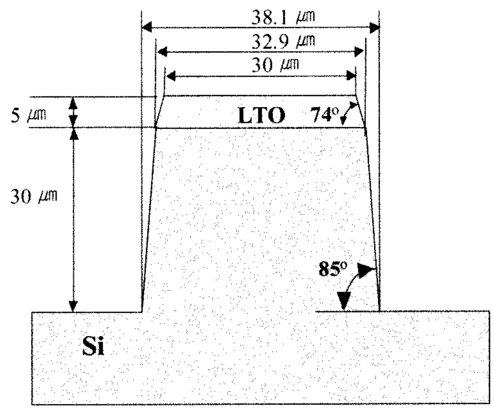
The probes were tested for their impedance and for their mechanical strength. A potentiostat (CH Instruments, model 660) was used for the impedance measurement. The peak-to-peak amplitude and the frequency of the ac voltage were varied over ranges from 10 to 270 mV, and from 1 Hz to 100 KHz, respectively. For mechanical testing, a stress test station (indenter) designed for testing microcantilevers (Akashi Co., model MZT4) was employed. While the PCB holder of the probe assembly was kept still, bending force was applied normal to the flat surface of the probe shank. The point at which the force was applied was kept at the same distance (100 μm) away from where the shank extends out of the unthinned carrier for all probes under test. The displacement was varied at a rate of 0.5 μm/s at the test point. The amount of force of the shank caused by the bending was recorded as a function of the displacement, thus a stress-strain relation [12] was derived for each shank thickness.

B. Animal Studies

Sprague-Dawley rats (200–300 g) were anaesthetized with urethane (1 g $\cdot$ kg^{-1}, i.p.). Animals were mounted in a stereotaxic frame (David Kopf) and craniotomy (2- to 3-mm diameter) was performed over the primary somatosensory (SI) cortex using the bregma as the initial point of reference. All procedures were done according to protocols approved by the Hallym University Campus Animal Care and Use Committee in order to minimize distress. Detailed methods of surgery can be found in [13]. The SNU probe was driven into the forepaw area of the SI cortex (0.5- to 1.0-mm anterial from bregma; mediolaterally, 3.5–4.5 mm; 0.5–1.2 mm deep from brain surface) with a micromanipulator. Stainless steel wire was used as a reference electrode, located in the lateral region of the brain. Cutaneous receptive fields were identified by listening to the recorded signal through an audio speaker while using a fine brush to tap the forepaw lightly, until the zone responding most intensely and reliably was defined. Electrical stimulation was provided by a bipolar concentric stimulating electrode (100-μ m tip, 0.5-mm tip separation, A-M System) and consisted of monophasic square pulses (pulse width 0.1 ms, frequency 1 Hz) passed through a stimulator (Model 1830, World Precision Instr.) with isolation unit to provide constant current (50–500 μA). The stimulating electrode was inserted under the center of the receptive field and was fixed firmly to prevent any movement. Responses were amplified for a gain of 1000 and bandpass filtered (homemade instrument) at 300 Hz to 3 kHz, passed



(a)



(b)

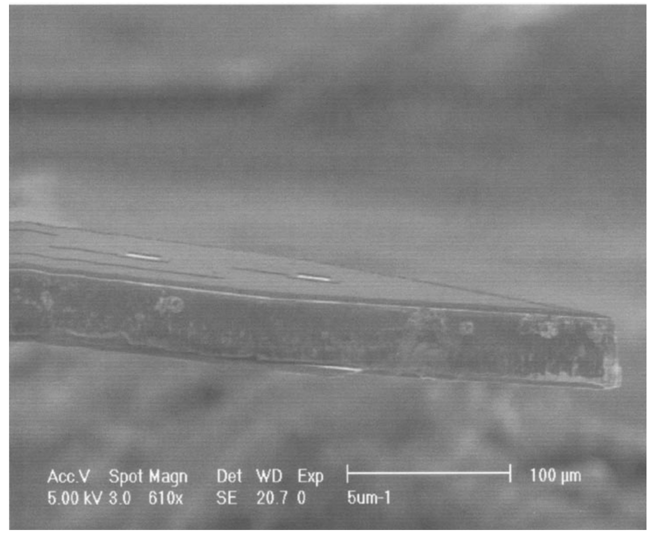
Fig. 2. (a) Scanning electron microscope view of a multishank probe after dry etching of silicon with Cl_2 gas. (b) Schematic drawing of a facet cross section in (a) with dimensions.

to a digital storage oscilloscope (Pro10, Nicolet Instr.) for visual inspection and then to the data acquisition system (CED 1401, Cambridge Electronics) for digitization (sampling rate 20 kHz, 12-bit resolution) and subsequent storage and analysis in a personal computer.

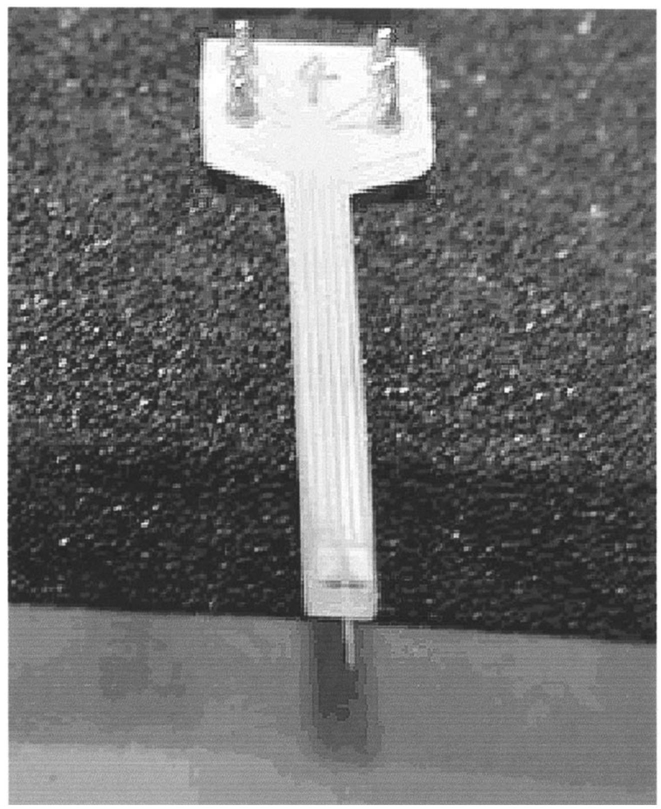
III. Results

A. Probe Fabrication

The range of the shank thickness obtainable by the LTO/dry etch process is estimated to be from 5 to 90 μm. The minimum thickness was determined by the KOH wet etching uniformity across four inch Si wafers. The maximum shank thickness was determined by the maximum LTO mask layer thickness and the dry etch selectivity. We could readily deposit up to 15 μm of LTO without cracking and the etch selectivity of Si over LTO was 6:1. The etch selectivity of LTO over Al mask layer was 16 to 1. Therefore a 1 μm thick Al mask layer could be used to define patterns on a 15 μm thick LTO layer, which then could be used to define 90-μm-thick shanks. The slopes of the dry etched LTO and Si walls were measured to be 74° and 85°, respectively. Fig. 2(a) and (b) shows the scanning electron microscope (SEM) picture of probe shanks after dry etching of Si to a thickness of 30 μm and the schematic view of a cleaved cross section, respectively. The pre-etch thickness of the LTO layer was 10 μm while its postetch value was 5 μm. Considering the slopes and the maximum layer thickness of LTO and Si, and starting with a 1-μm-wide Al pattern, we estimate that a shank 9-μm-wide and 90 μm high (an aspect ratio of ten) could be obtained.



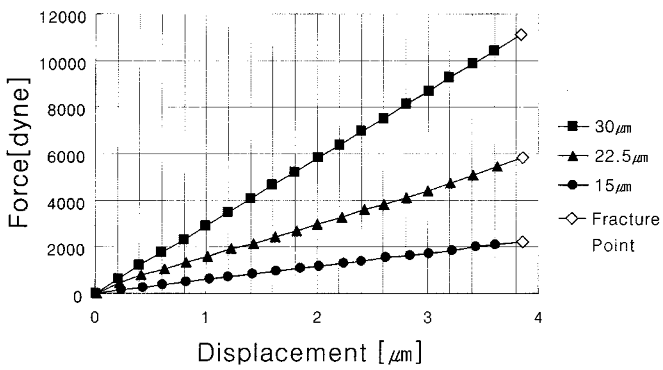
(a)



(b)

Fig. 3. (a) View of the probe tip using scanning electron microscope. (b) Probe packaged on a printed circuit board.

The SEM microphotograph of a 30-μm-thick, 200-μm-wide shank and the photograph of a probe bonded to a PCB are shown in Fig. 3(a) and (b), respectively. SNU probes with thickness of 15, 22.5, and 30 μm were tested for their mechanical strength.



(a)

Fig. 4. Results of the fracture strength test performed on probes with shank thickness of 15, 22.5, and 30 μm. The amount of force caused by bending was recorded as a function of displacement. The test was performed from one device for each thickness.

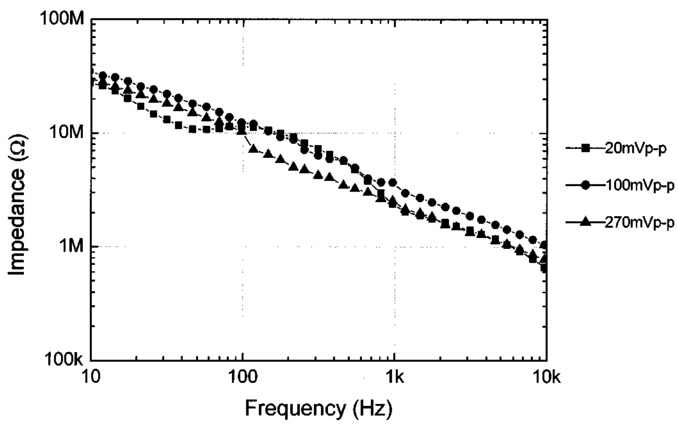


Fig. 5. Magnitude of the electrode impedance plotted against the frequency of the ac voltage source used in the potentiostat measurement. Three curves correspond to the three levels of peak-to-peak voltage used.

For each thickness, two probes were used in the test. Using the method described in Section II-A, we plotted the force as a function of the displacement as shown in Fig. 4. It is observed in the figure that the force-displacement very relation is linear up to the fracture point for all shanks. It is also noted that the amount of force which a probe shank can take increases for thicker shanks, while the amount of displacement (bending) for a given force increases for thinner shanks. The linearity is expected for silicon, being a glass-like material, in which plastic deformation is not present [12]. It is not clear, however, why all fractures occurred at nearly the same displacement and further study is needed. All of the fractures occurred at the carrier-shank junction. Fig. 5 shows the magnitude of impedance measured using the potentiostat mentioned above. The area of the electrode site was 600 μm^2, and saline was used as the electrolyte. The frequency of the ac voltage source was varied from 1 Hz to 100 KHz, while the peak-to-peak amplitude was set at 20, 100, and 270 mV. For the test voltage of 270 mV, 15 devices were tested. The average and standard deviation of the magnitude of impedance at 1 KHz were 2.1 and 0.6 MΩ, respectively. The dependency on the amplitude of the test voltage was not significant. At 1 KHz and 100 mV, the site impedance of the Ir electrode of Michigan probe with an area of 75 μm^2 was 3.0 MΩ, and that of the tungsten electrode (A-M Systems) with a 5-μm tip was 7.5 MΩ.

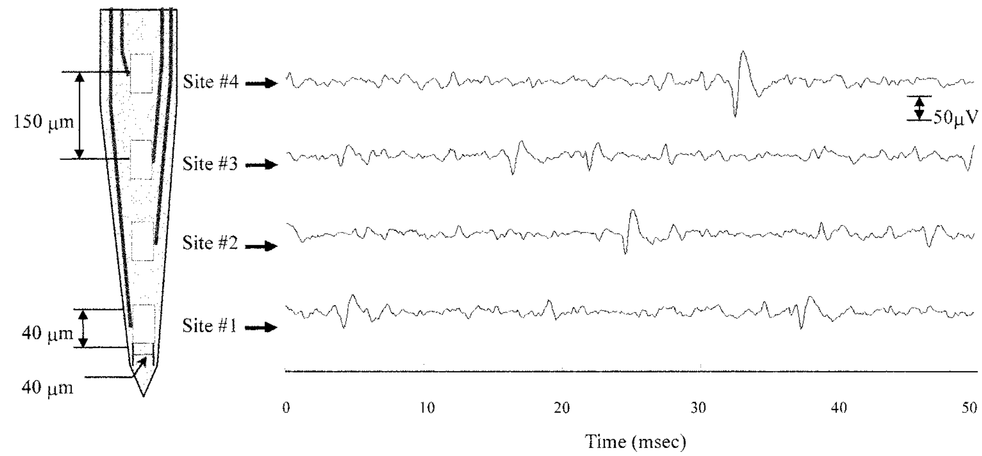
B. Recording Results

Simultaneous recordings of spontaneous neural activity from rat's somatosensory cortex obtained using the SNU probe are shown in Fig. 6. The instrumentation and animal protocols employed were as described in the Section II-B. Four sites, each with 1600 μm^2 area, separated by 150 μm from each other, were located perpendicularly through cortex as shown in the figure. The signals being recorded at the different sites are shown to be clearly independent. Each channel is distinguished from others by both the amplitude and the timing of the action potentials. The recording performance of the SNU probe was compared with that of the tungsten metal electrode and the Michigan probe in a separate animal study. The same region in the animal was probed by the three probes sequentially while the same instrumentation and animal protocols were consistently applied to all three electrodes. Signals from the SNU probe show comparable signal-to-noise ratios (SNR) to those measured using the other two probes. Typical SNR was 42, 37, and 40 dB for the SNU, the Michigan, and the tungsten probes, respectively. The SNR was defined as the ratio of the peak-to-peak amplitude of the largest action potential found in the recording interval to the rms value of the background noise in decibel. We have recorded action potentials for about five hours from each animal. During these periods, we did not observe any deterioration of the SNR in the recorded signals. No chronic recording has been attempted.

IV. Discussion and Conclusion

A new silicon depth probe was developed and its performance was evaluated through application to neural recording. Use of plasma etch with LTO masking allowed the thickness of the shanks to vary over a wider range than could be obtained by using the conventional boron diffusion technique. The range was estimated to be from 5 to 90 μm, determined by the back side wet etch uniformity (minimum thickness) and by the dry etch selectivity of Si over LTO (maximum thickness). Probes were constructed and tested whose thickness were 15, 22.5, and 30 μm.

The use of the LTO masking layer gives better uniformity and reproducibility because it involves only single coating and exposure of PR. LTO processing is available in many microelectronics fabrication facilities. Using the LTO mask, we could avoid problems associated with an alternative processing in which thick PR coats are used [10]. The latter process requires multiple coatings to create a thickness of more than 15 μm. This thickness is needed because of its poor selectivity over silicon in deep plasma etch. However, because conventional photolithography does not work with such thick coats of PR, it may be necessary to use several cycles of PR application and high energy exposure. These make it difficult to obtain uniformity and reproducibility. Additionally it is difficult to maintain high aspect ratio sidewalls in the thick PR so that dimensional control of the etched silicon is



(b)

Fig. 6. Four channel simultaneous neural recording from somatosensory cortex of rat. Each channel is distinguished from others by both the amplitude and the timing of the action potentials. The magnitude of impedance for the four sites was 2.4 ± 0.5 MΩ at 1 KHz.

compromised. As demonstrated in Fig. 2, with LTO, a well controlled high aspect ratio mask can be created. This control is also evident in the final etch characteristics of the silicon probe in the same figure.

Since this newly developed process does not involve high temperatures, it is compatible with standard CMOS processing. Thus the shank shaping process can be performed as a post process following the CMOS process. We can then use the CMOS foundry services such as MOSIS to fabricate circuits, before applying the micromachining process described in this paper. The conventional process using boron diffusion, however, does not allow this because the diffusion needs to be done at a higher temperature (15 hours at 1175°C to obtain a 15 μm depth [5]) than those used in most steps of the CMOS processing.

The results of the mechanical stress test add significance to the LTO/dry etch process. The test showed that more force can be applied with thicker shanks before breakage. Najafi and Hetke have concluded after applying buckling tests on their probes that a silicon shank with 15 μm could penetrate pia mater of guinea pig and rat, while a 30-μm-thick shank is needed to penetrate dura mater [14]. While the new LTO/dry etch method will etch 30-μm silicon in about one hour, the boron diffusion will take about 60 hours to reach the same depth. If even thicker shank is required, the etch time required in the new developed process increases linearly with required etch depth, but the diffusion time increases as a square function of the shank thickness required.

The animal experiments demonstrate that we have created functional probes which should be useful for acute animal studies. Simultaneous four channel recordings obtained from rat cortex showed clear independence among channels. The signals obtained using the SNU probes had similar electrical properties to those obtained using the Michigan probes and the tungsten probes, and were comparable to those reported using similar devices [10], [15].

While thicker probe was more preferable in mechanical strength, however, its damage to nervous system was not examined for comparison to thinner probes. It can be conjectured that the thinner probes cause less damage to tissue since the thinner probes occupy the less volume in tissue. However, the thinner probes become flexible while in tissue and can actually result in more damage. Histological studies are being planned to determine the optimal dimension of the shank.

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REFERENCES

[1] J. Kruger, "Simultaneous recording from many cerebral neurons: Techniques and results," *Rev. Physiol. Biochem. Pharmacol.*, vol. 98, pp. 177–233, 1983.
[2] D. Anderson, K. Najafi, S. Tanghe, D. Evans, K. Levy, J. Hetke, X. Xue, J. Zappia, and K. Wise, "Batch-fabricated thin-film electrodes for stimulation of the central auditory system," *IEEE Trans. Biomed. Eng.*, vol. 36, pp. 693–704, July 1989.
[3] K. Takahashi and T. Matsuo, "Integration of multi-microelectrode and interface circuits by silicon planar and three-dimensional fabrication technology," *Sensors and Actuators*, vol. 5, no. 1, pp. 89–99, 1984.
[4] W. Rutten, T. Frieswijk, J. Smit, T. Rozijn, and J. Meier, "3D neuro-electronic interface devices for neuromuscular control: Design studies and realization steps," *Biosensors and Bioelectronics*, vol. 10, pp. 141–153, 1995.
[5] K. Najafi and K. Wise, "A high-yield IC-compatible multichannel recording array," *IEEE Trans. Electron. Devices*, vol. ED-32, pp. 1206–1212, July 1985.
[6] A. Hoogerwerf and K. Wise, "A three-dimensional microelectrode array for chronic neural recording," *IEEE Trans. Biomed. Eng.*, vol. 41, pp. 1136–1146, Dec. 1994.
[7] G. Ensell, D. Banks, D. Ewins, W. Balachandran, and P. Richards, "Silocn-based microelectrodes for neurophysiology fabricated using a gold metallization/nitride passivation system," *IEEE J. Microelectromech. Syst.*, vol. 5, pp. 117–121, June 1996.
[8] G. Urban, J. Ganglberger, F. Olcaytug, F. Kohl, R. Schallauer, M. Trimmel, H. Schmid, and O. Prohaska, "Development of a multiple thin-film semimicro DC-probe for intracerebral recordings," *IEEE Trans. Biomed. Eng.*, vol. 37, pp. 913–918, Oct. 1990.

[9] E. Maynard, C. Nordhausen, and R. Norman, "The Utah intracortical electrode array: A recording structure for potential brain-computer interfaces," *Electroencheph. Clin. Neurophys.*, vol. 102, no. 3, pp. 228–239, 1997.

[10] D. Kewley, M. Hills, D. Borkholder, I. Opris, N. Maluf, C. Storment, J. Bower, and G. Kovacs, "Plasma-etched neural probes," *Sensor Actuators A*, vol. 58, pp. 27–35, Jan. 1997.

[11] D. Edell, "A peripheral nerve information transducer for amputees: Long-term multichannel recordings from rabbit peripheral nerves," *IEEE Trans. Biomed. Eng.*, vol. BME-33, pp. 203–214, Feb. 1986.

[12] R. A. Flinn and P. K. Trojan, *Engineering Materials and Their Applications*, 2nd ed. Boston, MA: Houghton Mifflin, 1981, pp. P534–P550.

[13] H. C. Shin, H. J. Park, and J. K. Chapin, "Differential phasic modulatio of short and long latency afferent sensory transmission to single neurons in the primary somatosensory cortex in behaving rats," *Neurosci. Res.*, vol. 19, pp. 419–425, 1994.

[14] K. Najafi and J. Hetke, "Strength characterization of silicon microprobes in neurophysiological tissues," *IEEE Trans. Biomed. Eng.*, vol. 37, pp. 474–481, May 1990.

[15] K. Drake, K. Wise, J. Farraye, D. Anderson, and S. BeMent, "Performance of planar multisite microprobes in recording extracellular single-unit intracortical activity," *IEEE Trans. Biomed. Eng.*, vol. 35, pp. 719–732, Sept. 1988.

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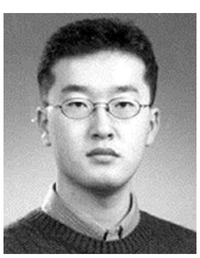

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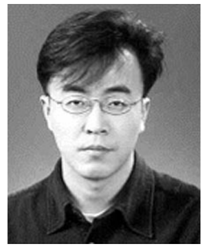

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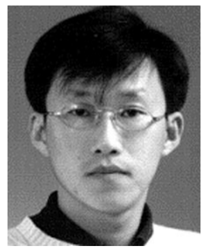

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