Abstract—This paper presents a novel approach to fabricate a high-density silicon microelectrode array on a silicon-glass bonded substrate. Electrodes were created with deep-reactive-ion etching (DRIE), and sharpened with a mixture of hydrofluoric, nitric, and acetic acids (HNA) wet etchant. A partial photoresist exposure method was developed to encapsulate the electrodes. A 10 × 10 microelectrode array was demonstrated in a 1 mm² area. In addition, a novel fabrication technique was incorporated to create the microelectrodes each with a different length to allow better access to the entire cross section of the target nerve bundle. Both the preliminary mechanical and electrical characteristics of the prototype devices were promising for use as peripheral intraneural probes.

Keywords-implantable electrodes; micromachining

I. INTRODUCTION

Direct stimulation of the peripheral nerves offers significant advantages over extraneural stimulation by providing increased spectral resolution and lower power consumption. There is an increasing demand for neural prostheses to restore sensory and motor function. Traditional single unit metal wire electrodes and glass capillaries are widely used in neurological studies. However, due to the bulky size of the electrodes and the complexity of peripheral nerves, it is not practical to implement neural prostheses by integrating single-unit electrodes.

With the advances of silicon micromachining technology that leverages heavily the lithographic techniques from the IC industry, drastic miniaturization of electrodes has been made possible. Silicon based microelectrode arrays have previously been fabricated with the intention of implantation in both the central and peripheral nervous systems by several research groups. One of the pioneering works reported was to use silicon surface micromachining and thin-film technology to fabricate two-dimensional (2D) multi-channel microelectrode arrays [1]. Several of these 2D probes could be micro-assembled into three-dimensional (3D) arrays [2]. Bulk micromachining has also been used to demonstrate silicon or silicon/glass based 3D arrays of microelectrodes for cortical recording [3] – [5]. Alternative materials other than silicon or glass have been explored as well, including electroplated nickel and NiFe (Permalloy) [6].

Nevertheless, these state-of-the-art research electrode arrays were normally fabricated with a shank-to-shank distance of more than 200 μm, resulting in limited number of channels in one array. Thus, they are not suitable for peripheral intraneural application. In this work, we reported a novel approach to fabricate high-density microelectrode arrays for peripheral intraneural applications (Fig. 1). The silicon based microelectrodes were created with bulk micromachining the silicon-glass bonded substrate and then sharpened with a wet etch process. This fabrication approach results in devices with little or no residual stresses, in contrast to a thin-film deposition process. The electrode tips were coated with platinum thin film with a partial exposure method, which formed the metal-electrolyte interface. A layer of poly(parachlororoxylene) (Parylene-C) covered the entire array except the tips of the electrodes to serve as an insulation layer. Each individual electrode can be connected to circuitries through the backside through-wafer metal contact vias. A novel technique was integrated with this process to create a varying-length microelectrode array to allow maximum access to individual neurons in the entire cross-section of a nerve bundle.

II. BASELINE FABRICATION

The baseline fabrication process is shown in Fig. 2, in which an array of equal-length microelectrodes was created. The process began with anodically bonding a 500μm-thick, highly-doped silicon wafer to a same-size Pyrex glass wafer, which was then patterned with contact vias. An array of 500 μm-tall pillars was created afterward with deep-reactive-ion etching (DRIE) through the thickness of the silicon wafer, with each pillar isolated from each other and aligned directly on top of a via on the glass wafer. The pillars are then sharpened into needle shapes with a mixture of hydrofluoric (HF), nitric (HNO₃), and acetic acids (CH₃COOH) (HNA) wet etchant. The array is then coated with platinum (Pt) and insulated with Parylene-C. The final step is to expose the tips with partial oxygen plasma. The following details the processing steps.

A. Silicon-Glass Substrate Preparation

Fig. 2(a) illustrates the first step of anodic bonding between a 100-mm diameter, 500-μm-thick silicon wafer (p-type 0.01–
used. Parylene-C was shown to be one of the best insulators for biomedical applications due to its high electrical resistivity and demonstrated long-term biocompatibility [7], [8]. The top surface of the device was first treated with a dilute solution of organic silane to improve adhesion, followed by conformal coating of Parylene-C through thermal polymerization reaction (Cookson Electronics Equipment SCS/PDS 2010). The electrode array was patterned with partial exposure of AZ4620 in the same fashion as the tip metallization step. Reactive-ion etching with oxygen (Plasma-Therm) was used to etch the exposed Parylene-C until the underlying tips were cleared. This step was performed with forward power set to 150 W and chamber pressure at 20 mTorr. The observed etch rate was 2 μm·hr⁻¹ (Fig. 2(h)). Fig. 3 are the Scanning Electron Micrographs (SEMs) of the silicon pillars after DRIE and the sharpened electrodes. Fig. 4 are SEMs of the tips undergoing metallization.

**Figure 2.** Cross sectional illustration of the process flow to create an array of equal-length microelectrodes. (a) A highly doped silicon wafer (top) is anodically bonded to a Pyrex glass wafer (bottom). (b) Contact holes are etched through the Pyrex wafer. (c) DRIE of the silicon wafer to create pillars. (d) HNA etch timed to sharpen the silicon pillars into electrodes. (e) Spin on AZ4620 photoresist. (f) Partial UV exposure and development. (g) Evaporate platinum and lift off. (h) Coat with Parylene-C followed by oxygen plasma to expose the Pt-coated electrode tips.

**Figure 3.** Scanning Electron Microscopy (SEM) of microelectrode array taken at 30° tilt. (a) DRIE result. (b) Partial HNA etch result with bullet-shaped tips. (c) Needle-shaped electrodes at the completion of HNA etch. (d) Close-up view of electrode tips, demonstrating tip angles less than 20°.

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**C. Tip Metalization and Insulation**

To lower the resistivity between the electrodes and the target neurons, the sharpened electrode tips were coated with a layer of 1,000 Å of platinum (Pt) with a 100 Å-thick of titanium (Ti) as adhesion layer. A modified lift-off process was employed, in which a layer of thick AZ4620 PR was spun on the sample at 1,000 rpm for 30s, which covered the entire front side, immersing the electrodes (Fig. 2(d)). The Pt was then flood exposed with UV light for 3min, which caused only the top layer of the PR to crosslink. After development, the tips were exposed while the majority of the shanks were still covered with unexposed PR, as shown in Fig. 2(f). Ti and then Au were e-beam evaporated to the desired thicknesses followed by lift-off (Fig. 2(g)).

To achieve insulation of the electrode bases from surrounding tissues, a 2-μm-thick layer of Parylene-C was used. Parylene-C was shown to be one of the best insulators for electroplating another 8,000-Å-thick layer. The via pattern was formed by first evaporating a 2,000-Å-thick film followed by a layer of 1,000 Å of platinum (Pt) with a 100 Å-thick of titanium (Ti) as adhesion layer. A modified lift-off process was employed, in which a layer of thick AZ4620 PR was spun on the sample at 1,000 rpm for 30s, which covered the entire front side, immersing the electrodes (Fig. 2(d)). The Pt was then flood exposed with UV light for 3min, which caused only the top layer of the PR to crosslink. After development, the tips were exposed while the majority of the shanks were still covered with unexposed PR, as shown in Fig. 2(f). Ti and then Au were e-beam evaporated to the desired thicknesses followed by lift-off (Fig. 2(g)).

0.05 Ω·cm and a 100-μm-thick glass wafer (Corning Pyrex 7740). This glass wafer served as the insulating substrate that would later provide mechanical support to the array of individually isolated electrodes. As illustrated in Fig. 2(b), the first masking step was to etch 50 μm-diameter contact vias through the thickness of the glass with 49% HF. A layer of 1-μm-thick gold (Au) film was used as the etch mask, which was formed by first evaporating a 2,000-Å-thick film followed by electronplating another 8,000-Å-thick layer. The via pattern was transferred to this Au mask by wet etching through a thick photoresist (PR) layer (Clariant AZ4620). To prevent introducing pinholes on the glass substrate, this PR mask was left on until the HF etch is completed. The observed etch rate of about 5 μm·min⁻¹. Several factors had been considered when optimizing the etch recipe to achieve the desired etch result at a reasonable etch rate. Since each of the electrodes started from a 80-μm-diameter circular pattern and spaced 50 μm from neighboring ones, loading effects would cause the etch rate to decrease as the etching progresses. Also, near the end of the etch, care must be taken to not over etch when the glass material was exposed, which would otherwise cause severe “ footing” on the silicon pillars. After this DRIE step, the PR mask was then removed with oxygen plasma.

To sharpen the pillars, an HNA mixture with 70% HF, 20% HNO₃, and 10% CH₃COOH was used with an etch rate of about 5 μm·min⁻¹. To minimize tissue damage during implantation, tip angle should be smaller than 20°. It was found empirically that when the etch was performed at room temperature in a static solution, sharp tips with angles less than 20° could be achieved. Agitating the etchant would minimize the loading effect and would result in uniformly narrowing the silicon pillars with almost no sharpening.

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**Figure 4.** SEMs of the tips undergoing metallization.
III. VARYING-LENGTH ELECTRODE FABRICATION

The above process resulted in an array of 100 microelectrodes of equal length. After implantation into a nerve fascicle, the array would allow access to nerve fibers lying on a plane spanned by the electrode tips, which would be parallel to the length of the fascicle. As a result, many of the nerve fibers would be redundantly contacted by multiple electrodes on the same row along the length of the nerve fiber, while other fibers situated outside the plane spanned by the electrode tips could not be accessed. To minimize redundancy and to access nerve fibers from a broader cross section of the fascicle, it would be advantageous to create arrays of electrodes with different lengths. One such 10 × 10 electrode array was demonstrated with machining silicon to form electrodes varying in length from 0.5 mm to 1.5 mm spaced 400 μm apart [9].

We derived a novel technique to achieve electrodes with different lengths with only one masking and etching process. It was observed the diameters of the silicon pillars formed from DRIE can be manipulated to affect how they would be etched in the subsequent HNA wet etch step. Generally, the silicon pillars would be shortened upon prolonged exposure to HNA after being sharpened. Since it took longer for the pillars with larger diameters to be sharpened, the pillars with smaller diameters will be shortened after all the tips are sharpened. Using this technique, we varied the diameters of the circular patterns used in the DRIE step to achieve varying-length electrodes. Unlike the Utah Slanted Electrode Array (USEA) reported in [9], we demonstrated a varying-length array with symmetrical arrangement of electrode lengths, in which the shortest electrodes are located in the center of the chip and increased in length towards both opposite edges. This geometry would potentially provide better handling during the implant process and would improve mechanical stability afterward.

The process for creating a varying-length electrode array is similar to the baseline process, with the exception that the silicon pillars are of different diameters after the DRIE step, as illustrated in Fig. 5. The first-generation prototype electrodes varied in length from 100 μm in the center of the die to 400 μm at the edge (Fig. 6).

IV. CHARACTERIZATION

A. Preliminary Electrical Characterization

The impedance of between the electrodes and simulated body fluids was estimated by submerging the microelectrode tips portion in 0.9% saline solution at room temperature, connected to a function generator and a precision 2 MΩ resistor in series with a large platinum counter electrode in the solution. A 1 μA sine-wave current at 1 kHz was generated while monitoring the voltage drop across the precision resistor. The impedance from the counter electrode through the solution to the microelectrodes under test was evaluated to be 2 MΩ at 1 kHz. Further characterization is underway with inter-electrode impedance measurements after implantation into nerve tissues.
B. Preliminary Mechanical Characterization

Since single-crystalline silicon is a brittle material, it is important to understand the influence of electrode geometry on its mechanical robustness during surgical implantation and stability thereafter. An experiment to simulate surgical implantation was performed by pressing a prototype electrode array into a layer of poly-dimethylsiloxane (PDMS), which has a Young’s modulus ($E$) of 500 KPa and a Possion’s ratio ($\sigma$) of 0.5. No broken electrode was found after the electrode array was pulled from the PDMS sample (Fig. 7). Note that nerve tissue is more than an order of magnitude softer than PDMS, with an $E$ of 34 KPa and $\sigma$ of 0.34 [10]. Therefore, the electrode array as designed should satisfy the robustness requirement in real applications. Further testing on nerve tissues is underway.

Figure 7. Optical micrograph of the PDMS sample punctured by the electrode array. No broken electrode was found.

V. Conclusion

In this paper, we demonstrated a novel fabrication process for a high-density penetrating microelectrode array for peripheral nerve stimulation and recording. A 10×10 electrode array was created in 1-mm² area with a novel bulk micromachining technique. An array of microelectrodes with lengths varying from 100 μm to 400 μm was demonstrated. Preliminary testing for both the electrical and mechanical properties of the microelectrodes was performed. Further characterization in biological environment is underway, with the ultimate goal of applying the electrode array for peripheral nerve implantation applications. Another potential use of the device could be for cortical or deep brain implants by adjusting the lengths and densities of the array to suit the final applications.

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