FABRICATION OF IMPLANTABLE POLYIMIDE BASED NEURAL IMPLANTS WITH FLEXIBLE REGIONS TO ACCOMMODATE MICROMOVEMENT

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ABSTRACT

A unique structure for chronically implantable cortical electrodes using polyimide polymer was devised, which provides both flexibility between brain tissues and skull and stiffness for easy insertion. The fabricated implants are tri-shanks with 5 recording sites (20 x 20μm) and 2 vias per electrode of 40 x 40μm. Each recording site was connected to the external circuitry via a 15-channel connector, which is especially designed to facilitate processing of neural signals to the external circuitry. Measured impedance values are in ~2Mohm at 1KHz. For a 5μm thick silicon backbone electrode, the stiffness was improved 10 times larger than that of the electrode without silicon backbone layer. Stiff electrodes with 5μm and 10μm thick backbone silicon penetrated pia of rat without buckling.

INTRODUCTION

Since the first advent of the simple intracortical single microelectrode four decades ago, there have been many significant advances in neural implants to read electrical signals in living nerve cells, to stimulate a variety of biological tissues, and subsequently help scientists better understanding how the brain works [1].

Present microelectrodes available today allows multiple electrode sites, chronically implantable, stiff enough to be inserted through the pia or the dura without causing significant trauma, and enable to deliver liquids in precise quantities [2]. However, the neural electrodes have not yet demonstrated the necessary longevity required to support greater strides in the basic neurophysiological research or clinical neruroprosthetic fields.

The ASU group recently reported the fabrication of polyimide based multichannel intracortical Bio-MEMS interface and demonstrated the neural activity from the somatosensory cortex area of rat's brain [3]. Polyimide (PI) based intracortical neural implant is very attractive for reliable and stable long-term implant function, because it is flexible and biocompatible. Flexibility is highly desirable to minimize tissue damage caused by the micromotion between the brain and the implant. Unfortunately, there is a problem associated with the flexible polyimide electrodes. Due to the lack of stiffness, polyimide flexible electrodes buckle during the insertion and therefore, cannot penetrate through the pia or the dura. For the first time we present a new design for chronic intracortical electrodes, that will provide both flexibility, between the external and inserted portion of the electrode, and adequate stiffness for easy insertion.

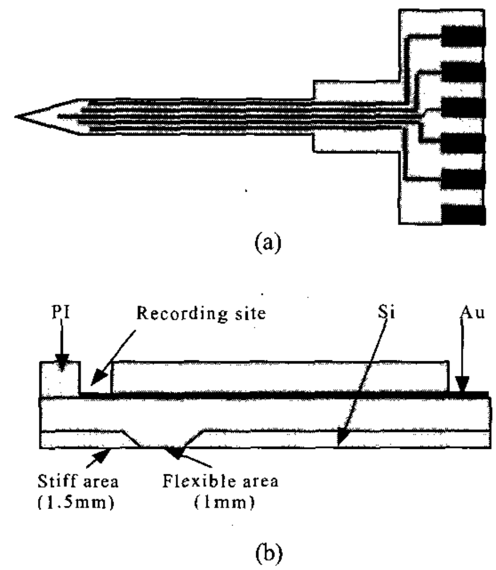

Figure 1 Simple schematic diagram of the polyimide based neural implant: (a) Top view and (b) cross-sectional view. The stiff portion with Si backbone layer remains inside the brain. Flexible portion without silicon backbone layer is to accommodate micromotion between brain tissues and skull. Each recording site is connected to the connector pads, in which the metal was exposed to atmospheres.

Fig. 1 shows simple schematic diagrams of the neural implant. For easy operation during surgical insertion, a 2~10μm thick silicon backbone layer, from SOI substrate, is attached to the desired region of the electrode to increase the stiffness (Young's modulus). The stiffness of the electrode can be varied by changing the thickness of the silicon backbone layer. The dimension of the stiff portion is 1.5 mm in length and 0.2 mm in width that remains inside the brain. It is then followed by 1 mm of flexible part of the electrode without silicon backbone layer designed to absorb stress from any micro-motion between the brain tissue and the electrode. Current microwire or silicon-based electrodes with stiff structure can cause additional damage to the surrounding neural or vascular tissues due to the relative micro-motion between the brain tissues and skull. Our new design can avoid these problems.

The recording sites are interfaced to the external circuitry via a 15-channel connector, which is especially designed to facilitate processing of neural signals to the external circuitry. The size of the connector portion is exactly same as a 15-channel connector.

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FABRICATION

Fabrication starts with a 4-in silicon-on insulator (SOI) substrate with varying device silicon thickness from 2 to 10μm and buried oxide thickness of 1μm. Top device silicon is (100) oriented *n*-type silicon with resistivity of 10~25Ω-cm. SOI wafer provides easy thickness control for stiff portion and excellent etching stop during backside etch process.

Fig. 2 shows schematic diagrams for fabrication procedure. Top device silicon layer was selectively etched away for flexible region using a 2000Å thick gold masking layer (Fig. 2(a)). To make smooth transition between flexible and stiff portions, silicon wet etching in 7% Tetra Methyl Ammonium Hydroxide (TMAH) was performed at 80°C. The silicon-etching rate depends on crystal planes in TMAH [4]. (100) crystal plane shows 20 times faster etching rate than that of (111) plane. Angled slope of the silicon removes step problem in next metallization process for recording sites (Fig. 3). After removing gold masking layer, the first layer of polyimide was spin-coated, exposed, and then developed as shown in Fig. 2(b). Then the polyimide layer was partially cured to protect the developed pattern from subsequent processing steps and provide a suitable surface for metal deposition. Partial cure of bottom polyimide layer and full cure of the upper polyimide layer terminate any route for water transmission through the boundary between the base and top polyimide layers.

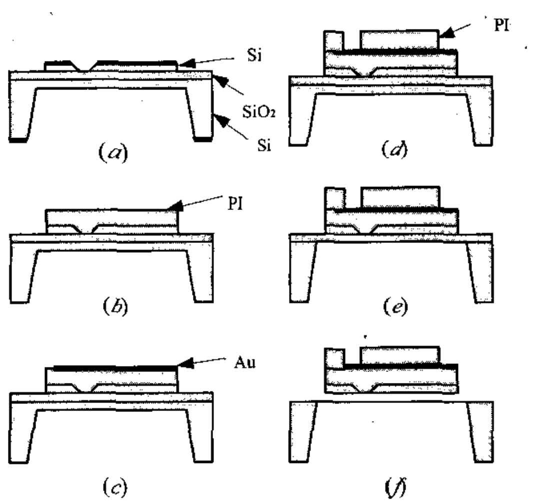

Figure 2 Fabrication procedure of the polyimide electrode with flexible region to accommodate micromovement. (a) Selectively etched top device silicon layer for flexible region using a gold masking layer, (b) the first layer of polyimide, (c) gold deposition for recording sites and gold traces, (d) the upper layer of polyimide to encapsulate or reveal the desired conducting surfaces, (e) backside silicon etching, and (f) buried SiO_2 etching and lifted final device.

A reactive ion etch (RIE) was used to clean and micro-roughen the polyimide surface prior to depositing the metal layers. After RIE, a 2000Å thick gold layer was deposited for recording sites, followed by wet etching (Fig. 2(c)). Gold was used for recording site because it has excellent surface inertness, and it provides no native oxide. However, gold is soft material, so long-term corrosion issues should be examed. The top polyimide layer was spun, exposed, and developed to encapsulate or reveal the desired conducting surfaces (Fig. 2(d)). The electrode was then fully cured to complete the imidization process. Polyimide shrinks vertically by about 20 to 30% during the curing process. The final polyimide structures are less than 20μm thickness.

Wafer was upside-downed for backside silicon etching in RIE. A top device surface was protected with photoresist and another dummy silicon wafer from plasma heat and RF power on the ground plane. Backside silicon etching was performed for 10hrs in RIE with SF_6 (25sccm SF6, 5sccm O_2, 100mTorr, and 120watt) [5]. Clean and uniform silicon backside etching was obtained (Fig. 2(e)). Silicon etching exactly stopped on the buried SiO_2 layer due to big etching rate difference between Si and SiO_2. After complete removal of backside silicon, the buried SiO_2 was etched away in 49% HF acid solution (Fig. 2(f)). Finally top-protecting photoresist was dissolved in microstrip for 2hr at 50°C. Several rinses with de-ionized water were performed to remove any unwanted etchant products.

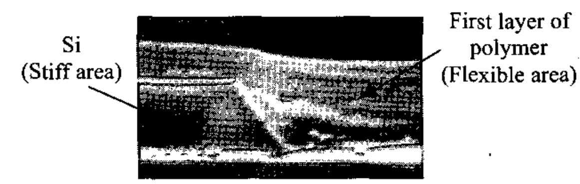

Figure 3 Cross-sectional SEM view of boundary between flexible and stiff regions. Angled slope of the silicon removes step problem in gold metallization for recording sites.

ELECTRICAL AND MECHANICAL RESULTS

The fabricated device was visualized through optical microscopy and scanning electron microscopy (SEM) as shown in Figure 4. The fabricated implants are tri-shanks with 5 recording sites (20 x 20μm) and 2 vias per electrode of 40 x 40μm. The stiff portion with silicon backbone layer is 1.5 mm in length and 0.2 mm in width that remains inside the brain. Flexible portion without silicon backbone layer is 1mm in length. The thickness was approximately 20μm. The connector portion of the completed electrode was exactly fitted into the commercial connector with 15-channel metal pads (Fig. 5).

Electrical test. Electrical impedance testing was performed using HP 4284A precision LCR meter. All the recording sites on the shafts were immersed into a 0.9% saline solution at room temperature [6]. Each recording site was connected to the external circuitry via a 15-channel connector. When an alternating test current source is passed through one of the recording sites and the saline solution, a potential develops at the electrode-electrolyte interface. This potential allows the recording site impedance. The area of the recording site was 400μm^2. 3 devices were tested. The frequency was varied from 100Hz to 10KHz.

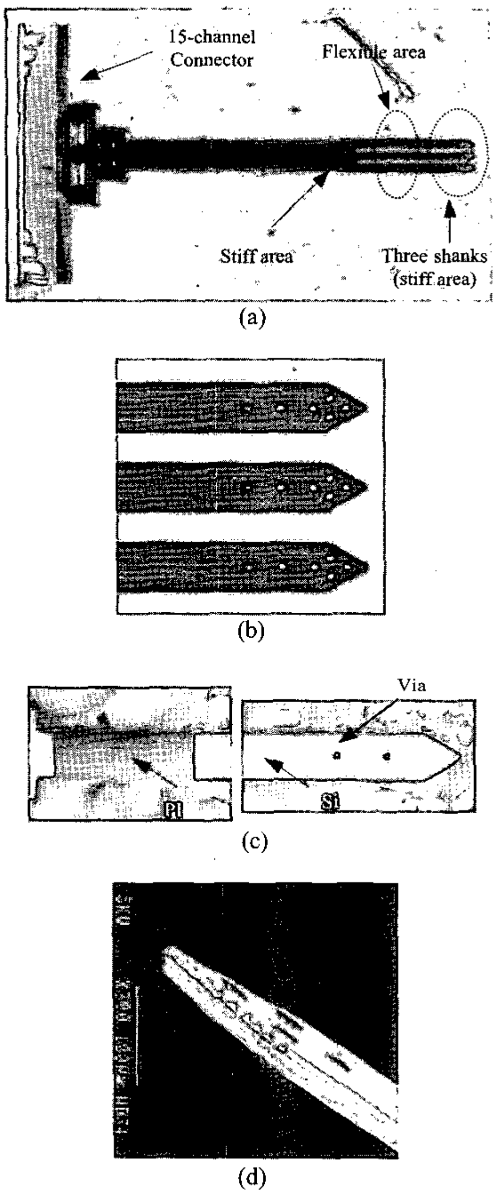

Figure 4. Optical microscope and SEM images of the fabricated electrode. (a) Entire view of the electrode, (b) top view of the three shanks, (c) backside view of the fabricated electrode, and (d) angled view of recording sites.

Fig. 6 shows the averaged impedance values from 3 devices at 100μA peak-to-peak current. The averaged impedance value at 1KHz was ~2 Mohm. The impedance remained stable over several weeks because of excellent water protection in the polyimide dielectric layers, and partial cure of bottom polyimide layer and fully cures of the upper polyimide layer. This curing approach terminates any route for water transmission through the boundary between the base and top polyimide layers.

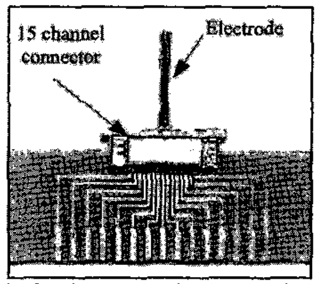

Figure 5 Optical microscope image of the connector and electrode. The connector portion of the completed probe was exactly fitted into the commercial connector with 15-channel metal pads.

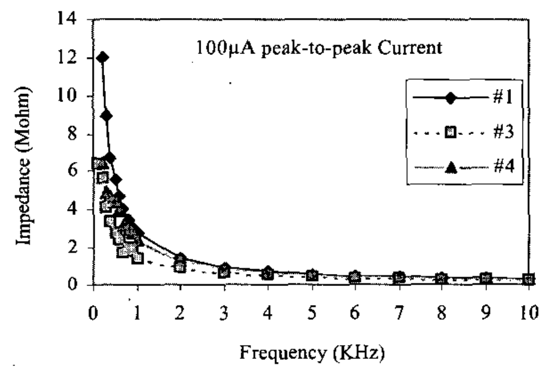

Figure. 6 Averaged impedance values from 3 devices. There are 5 recording sites per device. The recording site is 20 x 20μm and conducting pad is gold.

Mechanical test. Mechanical stiffness test of the stiff portion of the electrode shank was performed using micro force thermo-mechanical tester. Electrode stiffness is closely depending on the probe geometry such as width, length and thickness [7]. In a condition where all other parameters are same, the stiffness was tested in terms of the different silicon backbone thickness. The underneath backbone silicon thickness was varied from 2μm to 10μm. The probe dimensions are 1.5mm × 0.2mm × 0.02mm, in case where there is no silicon backbone layer.

The fabricated electrode was placed in the testing machine and a stretching force was applied until probe breaks. An extensometer was used to measure the amount that the electrode stretches between the gage marks. We convert the force to stress and the distance between gage marks to strain. The young's modulus is the slope of the stress-strain curve in the elastic region and the measure of the stiffness [8].

For a 5μm thick silicon backbone polyimide electrode, the stiffness was measured to be 31 Gpa. This value is 10 times · larger than that of the electrode without silicon backbone layer. It was further shown that the stiffness could be increased to 58 Gpa with a thicker layer (10μm) of silicon as shown in Table I.

The penetration test into rat's brain was performed to check whether the microprobe with silicon backbone layer could penetrate the pia and dura without any surgery aid tool (Fig. 7). For the test, rat was anaesthetized and heart rate and oxygen saturation were monitored. Skull and dura were removed and the stiff electrode was lowered to the surface (pia) by hand. Great care was made to encourage post implant recovery. Enough force was applied using Teflon tweezer. Stiff electrodes with 5μm and 10μm thick backbone silicon penetrated pia of rat without buckling. A 2μm thick silicon backbone electrode had fracture before creating an insertion into the pia (Table I).

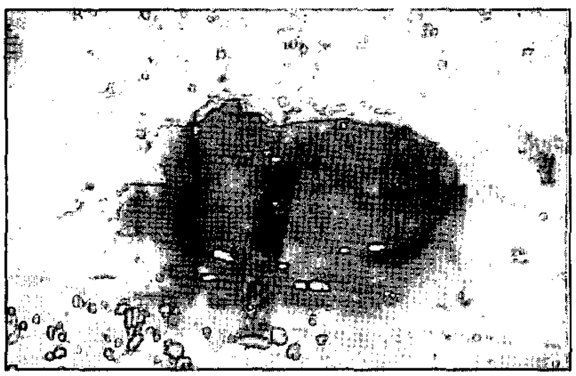

Figure 7 Optical microscope picture of the penetration test into rat's pia. A 5μm thick silicon backbone PI electrode is penetrating pia without bucking.

Table I. Measured mechanical stiffness values

	Young's modulus (GPa)	Rat Pia penetration
Pure PI electrode	2.8	No
2μm Si backbone electrode	10	No
5μm Si	31.8	Yes
10μm Si	58	Yes
Bulk Si	110	Yes

CONCLUSIONS

Polyimide polymer-based intracortical neural implant was fabricated for reliable and stable long-term implant function. Pure polyimide electrode had buckling problems during insertion. A new structure for chronically implantable cortical electrodes was devised, which provides both flexibility between brain tissues and skull and stiffness for easy insertion. For easy operation during surgical insertion, a 2~10μm thick silicon backbone layer, from SOI substrate, was attached to the desired region of the electrode to increase the stiffness. It was then followed by 1 mm of flexible part of the electrode without silicon backbone layer designed to absorb stress from any micro-motion between the brain tissue and the electrode. All the recording sites were positioned near the end of the shank in order to increase the probability of recording neural signals from a target volume of tissue. The area of the recording site was 400μm^2. The averaged impedance value at 1KHz was ~2 Mohm. For a 5μm thick silicon backbone electrode, the stiffness was improved 10 times larger than that of the electrode without silicon backbone layer. In penetration test into rat's pia, stiff electrodes with 5μm and 10μm thick backbone silicon penetrated pia of rat without buckling, while 2μm thick silicon backbone electrode had fracture before creating an insertion into the pia.

Acknowledgments

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