Benzocyclobutene (BCB) Based Intracortical Neural Implant

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Abstract

A novel structure for chronically implantable cortical electrodes using new Benzocyclobutene (BCB) bio-polymer was devised, which provides both flexibility for micro-motion compliance between brain tissues and skull and stiffness for better surgical handling. BCB is very attractive polymer for stable long-term implant function, because it has flexibility, biocompatibility, low moisture uptake (<0.2 wt%), and low dielectric constant (~2.6). For easy operation during surgical insertion, a 5~10µm thick silicon backbone layer is attached to the desired region of the electrode to increase the stiffness. 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. The fabricated implants have tri-shanks with 5 recording sites (20 x 20µm) and 2 vias (40 x 40µm) on each shank. BCB electrodes with 5µm and 10µm thick backbone silicon penetrated pia of rat brain without buckling.

1. Introduction

Recently the ASU (Arizona State University) group reported the fabrication of polyimide-based multichannel intracortical interface and recorded the neural activities from the auditory cortex of a rat's brain [1]. This type of polymer based intracortical neural implants present several attractive features: flexible, biocompatible, and easily manufactured using existing microfabrication technology. Flexibility is highly desirable to minimize tissue damage caused by the micromotion between the brain and the implant. Unfortunately, there are some problems associated with the flexible polyimide electrodes. Due to the lack of stiffness, flexible polyimide electrodes buckle during the insertion and therefore, cannot penetrate through brain tissues for neural implant application. In addition, polyimide has relatively high water uptake (4-6 wt%) and rough surface planarization.

Table 1. Electrical properties of BCB and PI

Properties	BCB	PI
Photo sensitivity type	Negative	Negative
Dielectric constant	2.65	3.5
Volume resistivity (ohm cm)	1×10^{19}	10^{16}
Breakdown voltage (V/cm)	3×10^{6}	3×10^{6}
Glass temperature (^{o}C)	>350	430
Water uptake (wt%)	0.12	4
Planarization level	Excellent	Poor

For the first time we report on the fabrication of benzocyclobutene (BCB) polymer-based intracortical neural implant for reliable and stable long-term implant function. BCB is a very attractive polymer, because it has flexibility, biocompatibility, low moisture uptake (<0.2 wt%), low dielectric constant (~2.6), and excellent chemical resistant [2]. In addition, the surface of the BCB is very amenable to modifications and preparations, which allow a host of bioactive organic species to be adsorbed or covalently bonded to its surface. Therefore, the BCB electrode will enable a new type of chronically neural implant for neural recording and stimulation. Some important properties of BCB and polyimide (PI) are listed in Table 1 [3].

Using CMOS-compatible processing techniques, BCB-based multi-channel neural implant was fabricated. Figure 1 shows a schematic diagram of the neural implant. Pure BCB polymer-based electrode had buckling problems during insertion. For easy operation during surgical insertion into neural tissues, a 5~10µm 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

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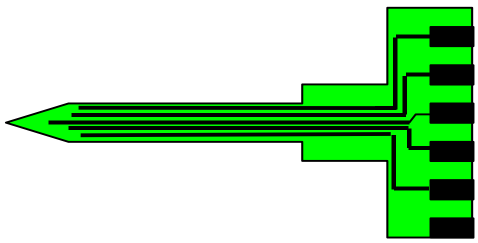
additional damage to the surrounding neural or vascular tissues due to the relative micro-motion between the brain tissues and skull. Our new design can avoid these problems.



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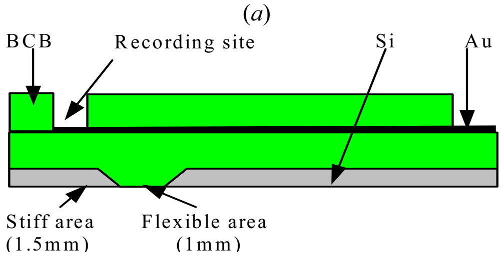


(*b*)

Figure 1. Simple schematic diagram of the BCB based neural implant: (a) Top view and (b) cross-sectional view.

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 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. The details are presented in this paper.

2. Fabrication

Fabrication starts with a 4-in silicon-on insulator (SOI) substrate with varying top 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.

Figure 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^{o}C. The silicon-etching rate depends on crystal planes in TMAH. (100) crystal plane shows 20 times faster etching rate than that of (111) plane [4]. Angled slope of the silicon removes step problem in next metallization process for recording sites, even BCB offers a degree of planarization of nearly 100% (Fig. 3). After removing gold masking layer, SOI wafer was cleaned and etched in an 80^{o}C, 4:1 solution of H_2SO_4 and H_2O_2. A reactive ion etch (RIE) was used to clean and micro-roughen the top silicon device surface prior to depositing the first BCB layer.

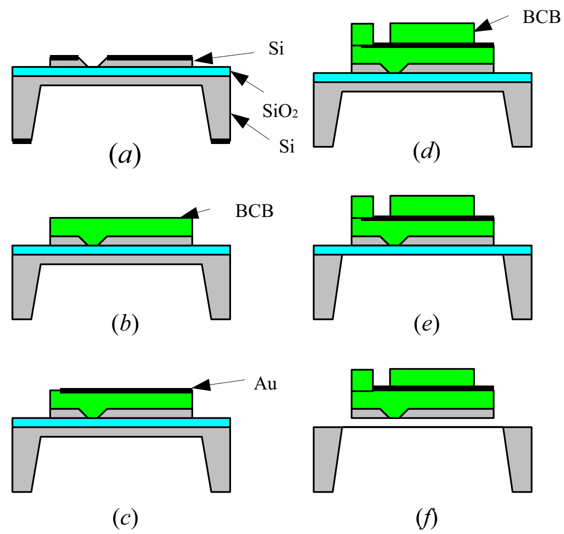


Figure 2. Fabrication procedure of the BCB electrode. (a) Selective etching of top device silicon layer for flexible region using a gold masking layer, (b) the first layer of BCB, (c) gold deposition for recording sites and gold traces, (d) the upper layer of BCB to encapsulate or reveal the desired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

The first layer of BCB (Cyclotene 4026^{TM} from Dow Chemical) was spin-coated, exposed, and then developed as shown in Fig. 2(b). Then the BCB layer was partially cured for 30 minutes at 200^{0}C in N_2 gas environment to protect the developed pattern from subsequent processing steps and provide a suitable surface for metal deposition. Partial cure of base BCB layer and full cures of the upper BCB layer terminates any route for water transmission through the boundary between the base and top BCB layers. Excellent planarization and small volume shrinkage on the base BCB layer were observed.

A reactive ion etch (RIE) was used to clean and micro-roughen the BCB surface. 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 BCB layer was spun, exposed, and developed to encapsulate or reveal the desired conducting surfaces

(Fig. 2(d)). The electrode was then fully cured for 1hour at 250^{0}C in N_2 gas environment. The final BCB structures are 20μm thickness. The recording sites were not completely opened, probably due to a lateral penetration of UV light. These residues hamper signal recording and electrical contact with a 15-channel connector. RIE was used to etch away the BCB residue on the gold surface using a 10μm thick photoresist-masking layer. A CF_4 and O_2 mixture was used for this descum process in RIE (100Watt, 100mTorr, 10sccm CF_4, and 40sccm O_2) [5].

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). 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 microstripe for 2hr at 50^{o}C. Several rinses with de-ionized water are performed to remove any unwanted etchant products.

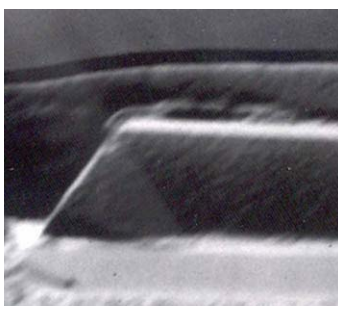
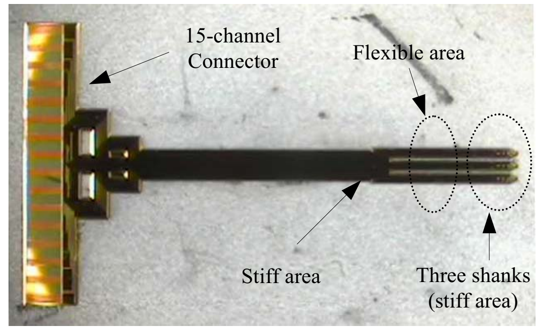


Figure 3. Cross-sectional SEM view of boundary between flexible and stiff regions. Angled slope of the silicon removes step problem in gold metallization for recording sites. BCB offers excellent planarization.

3. 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 (40 x 40μm) per electrode. 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).



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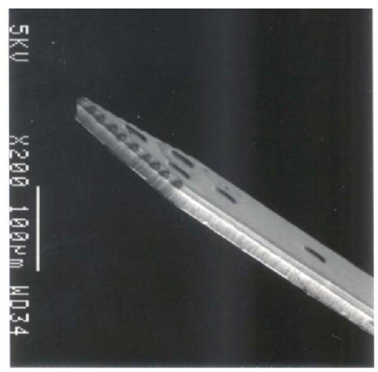

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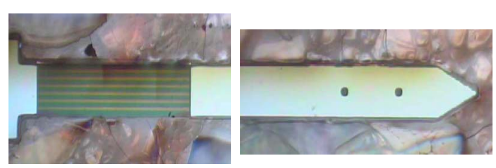

(d)

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

3.1 Electrical test

Electrical impedance testing was performed using HP 4284A precision LCR meter. All the recording sites were immersed into a 0.9% saline solution at room temperature [6]. When an alternating current source is passed through

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one of the recording sites and the saline solution, a potential develops at the electrode-electrolyte interface. This potential determines 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, while the alternating peak-to-peak current source was set at 50, 100, and 200µA.

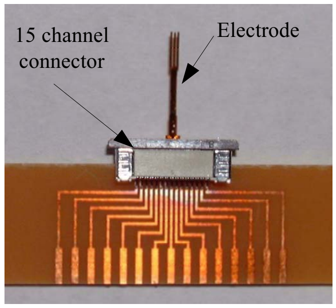


Figure 5. Microscopic image of the connector and electrode. The connector portion of the completed probe was exactly fitted into the commercial connector with 15-channel metal pads.

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 BCB dielectric layers, and partial cure of base BCB layer and full cures of the upper BCB layer. This curing approach terminates any route for water transmission through the boundary between the base and top BCB layers. Figure 6 shows impedance values depending on the peak-to-peak alternating test current sources. There is no any significant dependency on the testing current sources. The size and material of the recording sites mainly control impedance values.

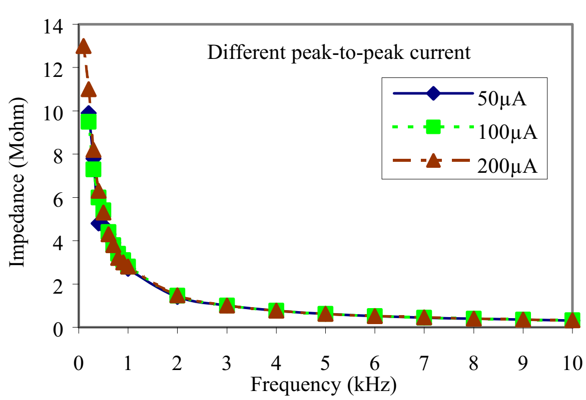
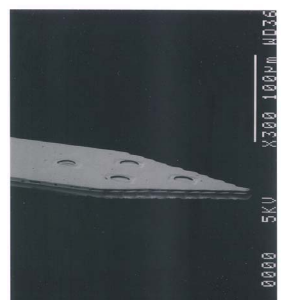


Figure 6. Measured impedance values in terms of frequencies. Three different peak-to-peak current were tested at one 20 x 20µm recording site.

3.2. Mechanical test

Mechanical stiffness test of the stiff portion of the electrode shank was performed using micro force thermo-mechanical tester. For this test, pure BCB and silicon electrodes were also fabricated, as shown in Fig. 7. Electrode strength and stiffness are 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 condition 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].



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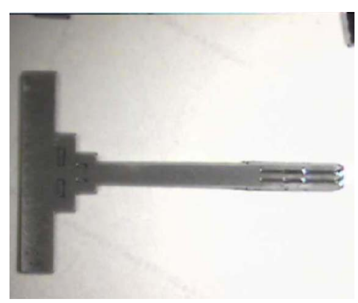

(b)

Figure 7. SEM images of (a) pure BCB electrode without silicon backbone layer and (b) pure silicon-based electrodes for penetration test.

For a 5µm thick silicon backbone BCB electrode, the stiffness was improved to 32Gpa. This increase is 10 times larger than that of the electrode without silicon backbone layer. It was further shown that the stiffness could be increased to 60 Gpa with a thicker layer (10µm) of silicon as shown in Table 2.

Figure 8. Optical microscope picture of the penetration test into rat's pia. 5μm thick silicon backbone BCB electrode penetrates pia without buckling.

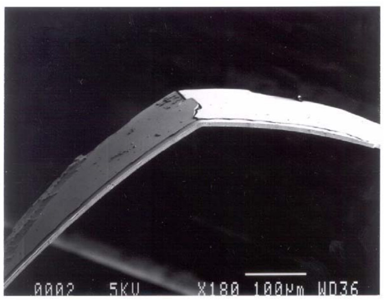

Figure 9. SEM picture with 2μm-thick broken backbone silicon. Electrode was fractured during insertion into the pia.

Table 2. Measured mechanical stiffness (Young's modulus) values

	Young's modulus (GPa)	Rat pia penetration
Pure BCB electrode	3	No
2μm Si backbone electrode	12	No
5μm Si	32	Yes
10μm Si	60	Yes
Bulk Si	110	Yes

Penetration test into rat's brain was performed to check whether the microprobe could penetrate the pia and dura without any surgery aid tool and buckling. 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 (Fig. 8). A 2μm thick silicon backbone electrode was fractured during an insertion into the pia, in which BCB film was not damaged, as shown in Fig. 9. Testing results are summarized in Table 2.

4. Conclusion

For the first time BCB polymer-based intracortical neural implant was fabricated for reliable and stable long-term implant function. Pure BCB electrode had buckling problems during insertion. For easy operation during surgical insertion, a 5~10μm thick silicon backbone layer was attached to the desired region of the electrode to increase the stiffness. 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. For the biocompatibility test, BCB was evaluated to be non reactive and non cytoxic. The area of the recording sites was 400μm^2. The averaged impedance value at 1KHz was ~2 Mohm. The impedance remained stable over several weeks because of excellent water protection in the BCB dielectric layers, and partial cure of base BCB layer and full cures of the upper BCB layer. 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 a 2μm thick silicon backbone electrode was fractured before creating an insertion into the pia.

Acknowledgments

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