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High-Yield Benzocyclobutene(BCB) Based Neural Implants for Simultaneous Intra- and Extracortical Recording in Rats

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***Abstract*—**A unique structure for chronically implantable cortical electrodes based on Benzocyclobutene (BCB) biopolymer was designed to perform intracortical and extracortical neural recording simultaneously in basic neuroscience research using animal models. It was fabricated on Silicon wafer using standard planar CMOS surface microfabrication technique. Dry-etchable BCB was used to insulate the electrode and provide flexibility for micro-motion compliance between brain tissue and skull. This electrode is also designed to ease the handling and implantation during the surgery and to integrate buffer circuits to improve the signal-to-noise ratio. The reliable fabrication process was developed to improve the electrode yield and performance. A 1.5um thick Tungsten layer was sandwiched in the electrode tip to improve the stiffness for easy insertion during the surgery. The fabricated electrodes have two intra-cortical recording sites (20×20um) in the tip penetrating into the cortex and two epidural recording sites (80×80um) on each side wing, providing a 6 channel system. One via (40×40um) was also incorporated in the tip to balance the tip and provide the bio-seeding to improve the implants and neural tissue interaction. The acute surgical testing suggests that this electrode structure can penetrate the pia into the cortical tissue without damaging the electrode.

Keywords: Biopolymer, Cortical recording, Neural Implant, Flexible electrodes, Integrated circuit

I. Introduction

Using silicon-based micro-machining technique, various recording and stimulating microelectrode arrays have been developed [1][2][3]. Though these penetrating electrodes can help us to understand how the neurons are exchanging information in the brain by detecting the neuronal activity in multiple locations, the current chronically implantable electrodes are designed and used in fundamental and applied neuroscience research in animal subjects whose movement and working environment are limited. Another type of brain research relies on electrodes placed epidurally or over the skull. This approach is less invasive but suffers from poor signal resolution and high noise effects. Since the intracortical action potential is strongly related to extra-cortical field potential, if the electrode is able to perform intracortical and extra-cortical recordings simultaneously, we can establish more realistic and robust mathematical relationship between these two potentials and realize the less and eventually non-invasive neural recording.

The recent work done at Arizona State University (ASU) demonstrated the fabrication process and applicability of photosensitive polymer based multichannel intracortical neural implants [3][4]. This type of electrodes presents several attractive features: flexible to accommodate micromotions or brain tissues and better tracking of individual neurons, engineered biocompatible by surface modifications, and perfect surface planarization. However, surgical handling and implantation of these electrodes are still difficult. Due to the high flexibility of polymers, even with 5um thick silicon backbone, the finished device usually curves due to imbalance of surface tension, which makes precise positioning and handling difficult. Also these electrodes have to be connected to the outside signal-processing unit through a microconnector that has to be hand connected during the surgery. The unavoidable moving or vibrating of the hand, though very little, will shake the electrodes to cause recording-sites deviation from the target neuron or neuron groups which will reduce the efficiency of the recording and the lifetime of the electrodes. Furthermore, the corresponding fabrication process was complicated especially with the silicon backbone adhered with the electrode, usually 6 masks were required and longer fabrication time are necessary to finish the whole process, hence the electrode yield in turn is relatively low and cost high. The last issue associated with these probes is the signal attenuation and noise coupling. Integrating the buffer circuits with the electrode can highly reduce noises and protect neurons from any possible electrical interference.

In this paper, we present a new electrode structure for simultaneously intra-cortical and extra-cortical neural recording that is designed to have better performance and long-term stable recording, as well as the corresponding optimized fabrication process for improving the electrode yield.

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II. STRUCTURES AND CIRCUIT DESIGN

Animal use in this research is reviewed and approved by ASU Institutional Animal Care and Use Committee.

Figure 1 shows the schematic view of the butterfly electrode structure. This structure consists of five parts. Part 1 is the penetrating tip which is designed to be 2mm in length to make sure the recording site goes to the fifth layer of the cortex. Two recording sites including one ground and one vias are incorporated inside the tip part. Each recording site is 20um×20um to provide enough impedance for recording of the neural signal and the vias (40um×40um) are incorporated for the purpose of seeding the bioactive components for improving the performance of the electrodes. Also this part has 1.5um tungsten sandwiched in the middle to make it stiff enough to penetrate through the pia. Part 2 is 10mm in length and 0.5mm in width; this part is totally flexible and can be bent up to 90 degree to help the positioning of the electrode tip and absorb the stress during the insertion.

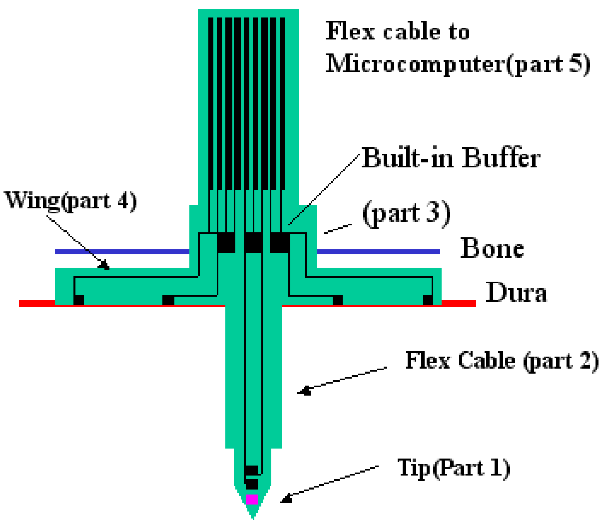


Figure 1. Schematic View of the BCB based butterfly electrode 6-channel structure with built-in buffer and flex cable

Part 3 is the headstage part, it is to be glued on the surface of the rat's head to ease the handling of the electrode, therefore, this part is designed to be 5mm in width and 3mm in length to meet the dimension requirement of the rat, later design for monkey subject can relax this requirement. Based on this dimension requirement, three LMC6035IBP low power 2.7V single supply CMOS Op Amp chips are used to form the buffer circuit (figure 2) to reduce the signal attenuation and noise coupling. Since the signal collected is typically as small as 50-500uV, this on-chip buffering can reduce the channel impedance and the parasitic shunt capacitance. These buffers are connected as voltage followers as shown in figure 3. It has a high-drive capability and low output impedance. Each buffer is 1.4mm ×1.4mm in dimension and contains two Op Amps. This part also has 1.5um W backbone underneath to avoid the problem during the soldering of the Op Amps. Part 4 consists of two wings; they are designed to perform the extra-cortical recording of the field potential. Two recording sites (80×80um) are incorporated inside each wing. The total length of each wing is 1.8mm. Part 5 is the flexible cable designed to connect the electrode to the ZIF connector that finally connect the electrodes to the computer for the signal processing. This part is 10mm in length and 4mm in width and it is totally flexible to avoid the moving of the headstage during the connection.

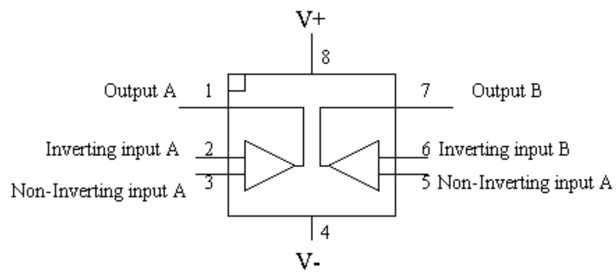


Figure 2. Schematic view of the single unit-gain buffer

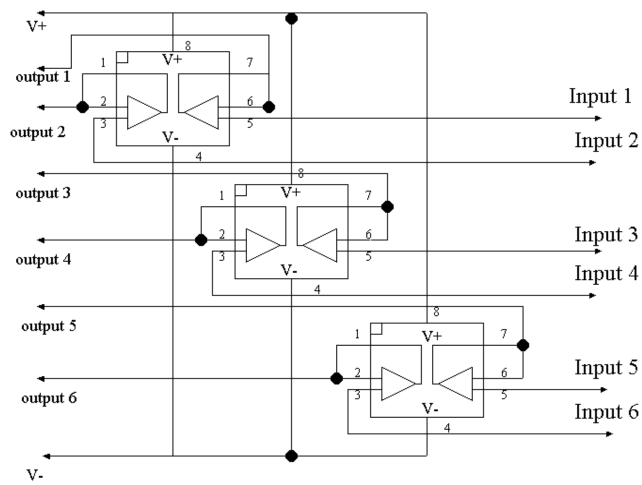


Figure 3. Schematic view of the head stage circuits

III. FABRICATION PROCESS

Figure 4 shows the process flow for this BCB based electrode. Start from 4-in (100) oriented n-type Silicon wafer with resistivity of 10-25 Ω-cm, 0.5um thick silicon dioxide was grown on the surface by wet oxidizing and the first layer of BCB (Cyclotene 3022 from DOW Chemical) was spin-coated at 1500 rpm to obtain about 20um thick BCB layer after curing. The thickness of the BCB is highly depended on the spin speed (Figure 5). Usually 20um thick BCB is needed to prevent the curving up of the finished probe. The sample was then partially cured for 40 mins at 210°C in N_2 gas environment to provide a suitable surface for metal deposition. Partially cure of base BCB layer and fully cure of the final 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 BCB layer was observed. A 1.5um thick Tungsten layer was sputtered onto the BCB surface (figure 4a) to improve the stiffness of the electrode tip and headstage area followed by the wet etching. 2000Å gold was used as the wet etching mask and was etched away after the W wet etching. Traditional photoresist cannot be used for the etching mask since it erodes at high temperature[5]. Due to the strong stress during the deposition, tungsten layer with thickness over 1.5um will crack the bottom BCB film. However, since the young's modulus of the tungsten is about 4 times bigger than that of silicon, 1.5um is enough for necessary stiffness enhancement.

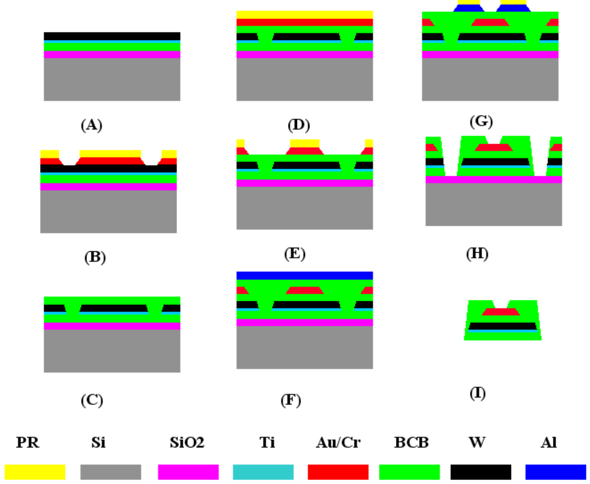


Figure 4. Process flow of the BCB based electrodes

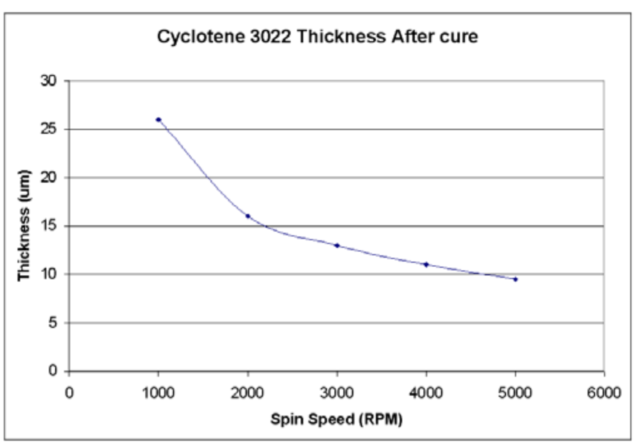
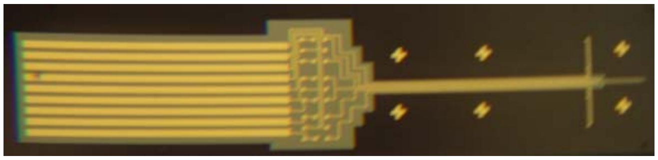


Figure 5. Thickness of BCB layer versus spin speed

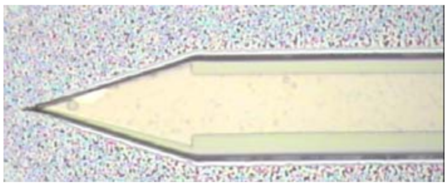
The second BCB layer was spin-coated and partially cured to completely encapsulate the tungsten layer from the conduction layer. 2000Å thick gold layer was deposited for recording sites, and conduction traces followed by wet etching (figure 4b, 4c). Gold was used for recording sites because it has excellent surface inertness and it provides no native oxide. However, gold is soft material, so long term corrosion issues should be examined. The top BCB layer was spin-coated and partially cured for 40 mins at 210°C in N_2 gas environment to encapsulate the whole device. After that, 3000 Å Al was E-beam deposited onto the surface as the BCB dry etching mask. (Figure 4f) Positive photoresist was coated and patterned as the Al wet etching mask. And the aluminum was wet etched to define the device shape. The BCB was then dry-etched by O2/CF4 plasma (figure 4h) in reactive ion etch mode (RIE, 100mTorr, 25sccm CF_4, 100sccm O_2, 150Watt). In RIE, reactive species are accelerated toward the surface, which results in a more anisotropic etch. Moreover, physical sputtering and chemical reaction combination can give higher etch rates. Unlike most other polymers, BCB requires F radicals in addition to O radicals for etching because of the presence of Si in the polymer matrix. This precludes the use of PECVD SiOx or SiNx as etch masks [6]. In this process, additional etching was performed to reveal the desired conduction surface and the recording sites. After that, the aluminum was removed and the samples were fully cured for one hour at 250°C in N_2 gas environment to covert 90% BCB to polymer. 49% HF was used at last to etch away the bottom silicon dioxide layer, and the device was lifted off the silicon substrate (figure 4I).

IV. RESULTS AND DISCUSSION

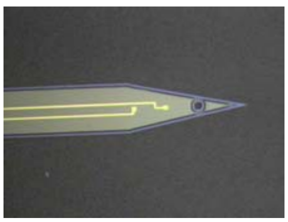
The Fabricated device was visualized through optical microscopy and X-ray microscopy as shown in figure 6. Figure 6a shows the overall view of the butterfly electrode and figure 6b shows the electrode tip with 1.5um tungsten sandwiched in the middle. Two intra-cortical recording sites (20×20um) and one via (40×40um) were incorporated inside the penetration tip portion (2mm in length and 0.15mm in width, figure 6c). And two extra-recording sites (80×80um, figure 6d) were incorporated inside each wing region (1.8mm in length and 0.2mm in width). The flexible cable connecting the front tip with the headstage is 10mm in length and 0.6mm in width. The other flexible cable connecting the headstage to the outside microcomputer is 10mm in length and 5mm in width with the end part exposed for connection with micro-connector. These two cables are totally flexible without tungsten backbone. The finished electrode has total thickness of about 50um and showed no curve-up problem. Figure 6e shows the headstage with three buffers soldered. The total fabrication time is only one week, and the electrode yield is close to 95%.



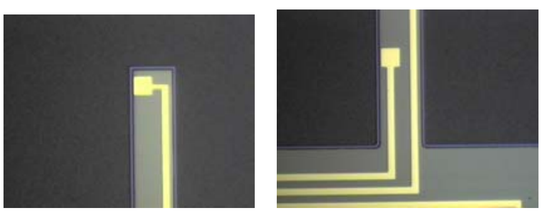
(a)



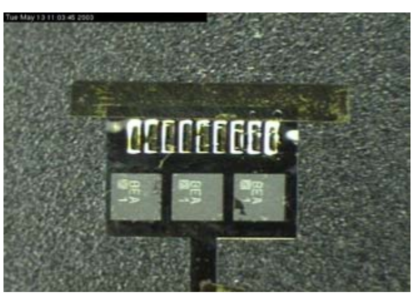
(b)



(c)



(d)



(e)

Figure 6. Optical microscope and X-ray images of the fabricated butterfly electrodes. (a) Overall view of the electrode. (b) Penetrating tip with 1.5um thick tungsten sandwiched in the middle of two BCB layer. (C) Top view of the tip showing the gold trace and via. (d) Top view of the wing region with two extra-recording sites. (e) X-ray image of the headstage with 3 Op Amps soldered.

Penetration test into rat's brain was performed to check whether the probe could penetrate into the pia without any surgery aid and buckling. The rat was anaesthetized and heart rate and oxygen saturation were monitored. Skull and dura were removed and the electrode was bended to the surface (pia) by hand. Great care was made to encourage post implant recovery. Enough force was applied using Teflon tweezer. Test showed that the electrode with 1.5um thick tungsten backbone penetrate the pia of rat without buckling.

V. CONCLUSION

In conclusion, a new neural probe aimed to realize intra-cortical and extra-cortical neural recording was designed and fabricated for reliable and stable long-term implant function. Compared to BCB electrodes with silicon backbone, fabrication process was optimized and the fabrication time was reduced which in turn improved the fabrication efficiency. For easy operation during the surgery, new headstage structure was designed and 1.5um thick tungsten was incorporated in the middle. Two flexible BCB cable was integrated with the electrodes to absorb the stress from any micro-motion between the brain tissue and the electrode. Three buffers were soldered on the headstage to improve the signal-to-noise ratio. The penetration test shows this electrode can easily and safely penetrate into the pia without buckling.

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