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A Cortical Recording Platform Utilizing μECoG Electrode Arrays

Jiwan Kim, J. Adam Wilson, and Justin C. Williams, *Member, IEEE*

Abstract— **Clinical applications of brain implantable devices for recording and interpreting electrical signals from the cortex have grown rapidly in the last decade. For long-term cortical recording, a micro-electrocorticographic (μECoG) electrode and universal platform were developed and evaluated. The electrode diameters and inter-electrode distances of the new device are on the order of 100s of μm, significantly smaller than general ECoG grids, and do not require penetrating the brain. Acute recordings from the device demonstrated that independent brain activity could be recorded from electrodes with a spatial resolution of 1 mm.**

I. Introduction

Recently brain-computer interface (BCI) technology, particularly with regards to neuroprosthetics, has become a common area of interest in the neuro-engineering realm [1]. Accelerated interdisciplinary research has been achieved by the collaboration between neuroscience, medicine, and rehabilitation engineering, bringing the common goal of technology for treating severe motor impairment closer to reality. This endeavor has also shown potential in the elucidation of the neural mechanisms inherent in the central nervous system [2]. Devices for brain signal acquisition, essential components of BCI systems, have lead to a plethora of novel bio-signal acquisition modalities [3-5]. Issues of compatibility of the device in physiological environments, as well as feasible fabrication techniques for making smaller feature sizes are key requisites of these devices.

The Electroencephalogram (EEG) has been the primary focus of clinical BCI technology thus far, due to its safety and convenience. However, EEG signals lack the resolution, amplitude, and bandwidth offered by more invasive methods [6]. Various invasive microelectrode arrays such as Michigan probes and the Utah electrode arrays have been developed within the last 20 years to achieve more localized neural signals [3-5]. Since these devices are based on silicon microfabrication technology, with micron scale features, they are capable of detecting single neuron activity. Though their capability to detect minute changes in action potentials is very impressive, inserting the probes (arrays) into brain tissue can cause significant scar tissue accumulation around the device (cell encapsulation/ensheathing), which may decrease the intensity of signal and the signal to noise ratio (SNR), and can lead to a loss of recording ability [7].

Recent Electrocorticogram (ECoG) studies have shown promise for its use as an alternative method for BCI control [8, 9]. Unlike invasive electrodes requiring penetration into the cortex, ECoG electrodes are placed on the cortex surface, which can reduce risk of possible implant related damage [10, 11]. ECoG also has some advantages over EEG including higher spatial resolution, broader bandwidth and higher amplitude than EEG. Recent studies have shown that ECoG signals can be used for BCI control with minimum training and they contain detailed aspects of motor actions, which was previously thought to be only possible using invasive electrodes [8, 9, 12].

A flexible and scalable polyimide μECoG device is presented in this report. In recent results, independent signals over a wide range of frequencies were detected using a common ECoG grid with 5 mm electrode spacing [9]. μECoG electrode arrays were designed on the order of 100's of microns, because the amount of information extracted using ECoG could potentially be improved by increasing the density and number of electrodes. This smaller μECoG electrode array also could be used for recording in the central sulcus, where a large area of motor cortex lies, in the anterior bank of the central fissure. The size of the burr hole required for the implantation of a μECoG system is significantly smaller than required for traditional ECoG grids, and does not penetrate the brain.

II. Materials and Methods

A. Microfabrication of μECoG Electrode Arrays

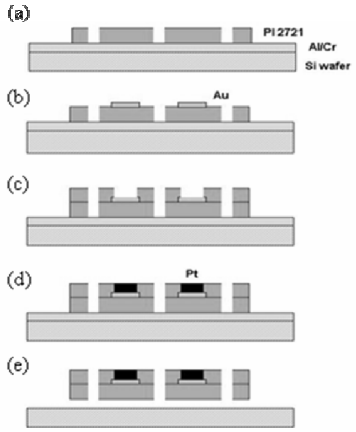

Fig. 1. Microfabrication process of μECoG electrode arrays; (a) Photo-patterning of polyimide on Al/Cr coated Si wafer (b) A lift-off process after Au metallization (c) Photo-patterning of polyimide as top layer (d) Electroplating of Pt black (e) Substrate release by metal dissolution technique.

Polyimide PI-2721 (HD Microsystems) was chosen over other polyimides to fabricate flexible μECoG electrode arrays

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Authors are with the Biomedical Engineering Department, University of Wisconsin, Madison, WI 53706 USA (e-mail: jwilliams@engr.wisc.edu).

because of its excellent insulation characteristics and because it can be microstructured by photolithography. The multilayer process for integrating metal tracks, micro-vias and interconnection pads is illustrated in fig. 1. For release of the microstructures, chrome (10nm) and sacrificial aluminum (300nm) were evaporated on 100 mm silicon substrates. On the aluminum surface, a 20 µm thick layer of PI-2721 polyimide was applied, photo-structured and cured for the bottom layer in a nitrogen atmosphere oven. The metallization (10 nm Cr/250 nm Au) for interconnection pads, connecting lines, and electrode sites was deposited using e-beam evaporation. Structures were patterned applying a lift-off technique. After metallization, a second polyimide layer for insulation was spun onto the wafer and photo-defined. Platinum black was electroplated onto the microelectrodes sites to reduce their interfacial impedance.

The µECoG electrode arrays were released by anodic metal dissolution technique. The wafer with the substrates to be release is immersed in a concentrated sodium chloride solution together with a Pt counter electrode. A positive 0.8 V is applied to the aluminum layer using a constant-voltage supply. During the anodic metal dissolution process of aluminum, the chrome remains on the substrate due to the difference in electrochemical potential compared to aluminum. After their release, the µECoG electrode arrays are rinsed in DI water and subsequently dried.

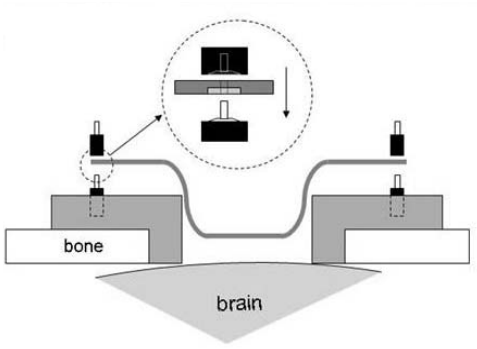


Fig. 2. Schematic diagram of housing platform and snap-in connectors.

B. Fabrication of a Housing Platform

A thermoplastic polyetherimide (PEI) (Ultem 1000, GE Plastics) was chosen as the housing platform of the initial proof-of-concept device. PEI has a stable dielectric constant, making it an ideal choice for electrical housings. In addition, PEI has outstanding mechanical properties, broad chemical resistance, and is generally regarded as bio-inert [13].

A schematic of the housing platform for the µECoG electrode arrays is shown in figure 2. A solid rod of PEI was machined to the corresponding dimensions (19mm hole and 3mm thickness) of the removed plug from the subject animal's skull. It has a flat region for the device attachment screws, electrical connectors, and future on-board circuitry.

Flexible devices are so thin (30µm), it is very difficult to find proper connectors that are feasible to fabricate, while maintaining their reliability. Figure 2 also describes the mechanism of snap-in connectors included in the housing platform. An identical 10 pin connector is snapped over the base mounted 10 pin array sandwiching the individual electrode sites between the two arrays. It is a press-fit type interconnection which doesn't require any soldering.

C. Acute Procedure of in vivo Chronic Testing

The µECoG electrodes were tested in an acute surgical procedure in rhesus macaque primates. The monkey was intubated and maintained on isoflurane for the duration of the surgery. Following anesthetization, the masticatory and cranial muscles were reflected to allow an exposure of the scull approximately 20 mm x 20 mm. A 19 mm trephine blade was used to drill a hole in the skull over sensorimotor cortex unilaterally, according to stereotactic coordinates. The µECoG device containing external connectors was inserted into the hole. The dura mater was left intact for epidural recording.

Signals were recorded with a Tucker-Davis Technologies RX5 (TDT, Alachua, FL) at a sampling rate of 24,414 Hz, band-pass filtered from 1 to 400 Hz with a 60 Hz notch, and down-sampled to 1061 Hz. Spontaneous brain activity was recorded from epidural surface for a total of 10 min. with a isoflurane level of 1.5%. The isoflurane level was then lowered to 0.5%, and brain activity was recorded epidurally for another 10 min.

III. Results and Discussion

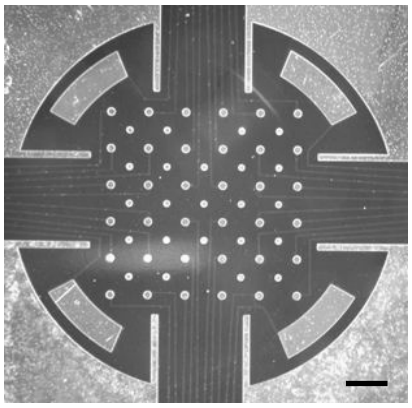
A. Fabrication process of µECoG Electrode Arrays

Neural implants have to fulfill complex requirements for long-term implantation. They have to be very stable in the physiologic environment and also flexible to conform to the soft, convoluted structures of the brain. Among other polymers, photodefinable polyimide PI 2721 was particularly chosen because of high resolution and various range of thickness (over 20µm). It exhibits excellent insulation characteristics and good chemical/thermal resistance as well as high flexibility [14].

As seen in figure 3(a), µECoG electrode arrays consist of electrode sites and holes, which are intended to facilitate consistent perfusion of brain for a long term implantation. The presented µECoG electrode is designed for the brain size of a monkey. The relatively large brain of a monkey allows for more variation in electrode structure. Four large corner electrodes were included as reference electrodes, increasing the recording isolation of the device. Figure 3(b) shows the completed assembly of µECoG electrode arrays and a housing platform before connecting cables. With all four arms bending, a clearly flat plane containing the electrodes is observed. The gaps between the electrode arrays (including electrodes and holes) and arm (including electrical lines) help make a full contact with brain surface. The µECoG electrode arrays are not intended to penetrate brain tissue, but to gently contact the brain surface. The assembly can be used as a

universal platform, accommodating a standard 19mm burr hole through the skull. During *in vivo* tests, we observed that surface of the μECoG electrode arrays maintains a sticky contact with the brain surface, and that four arms containing electrical leads buckle naturally. This buckling would provide an advantage in long term implantation. Buckling arms absorb micro-vibration in all three axes and the recording device remains stable even when brain pulsing occurs.

(a)



(b)

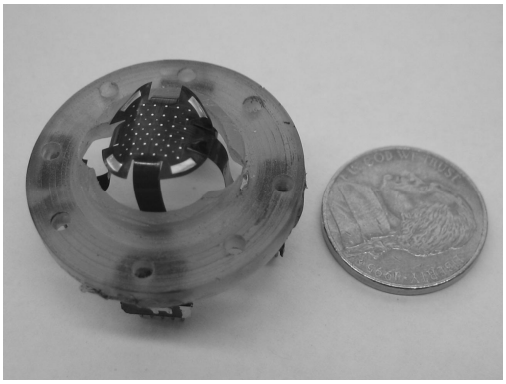


Fig. 3. (a) μECoG electrode arrays for monkey brain. Scale bar is 1mm. (b) Assembly of μECoG electrode arrays and a housing platform. A nickel was placed next to the platform to compare the size.

The microfabrication process for the μECoG electrode arrays did not include any wet or dry etching steps. Traditionally a wet etching method using a HF solution is utilized to release completed structures from oxidized silicon wafers. This technique may lead to over-etching and critical damage to the device. Since releasing the device is usually the final step, small damages can jeopardize the integrity of the final device. Dry etching by chlorine gas is also a common technique to make via holes in polyimide, but requires an additional mask set. Moreover, most gases used in dry etching processes, including chlorine, are very harmful to biological tissues. Any dry etching process residue, even on the order of parts per million, could be potentially injurious to biological systems, compromising the biocompatibility of the device. Besides, the metal dissolution technique for releasing completed devices is a gentle and safe process [15]. Release is carried out at room temperature and devices are smoothly released and float in the solution when it is finished.

One of the most useful techniques in our microfabrication process is rapid prototyping. Prototyping of masks using a high-resolution printer is a convenient method in biological laboratories as they allow for the fabrication that does not rely upon high-resolution chrome photomasks [16]. It is also a very flexible technique that expedites the implementation of new designs and decreases overall design cycle time.

B. Electrical Characteristics

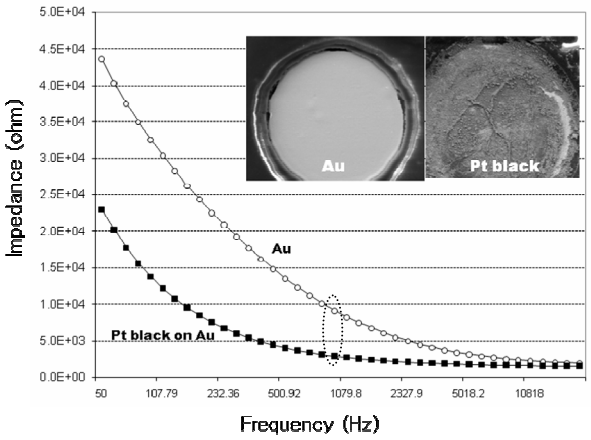


Fig. 4. Impedance measurement of gold and platinum electroplated electrodes. Dashed oval indicates the data points at 1 KHz.

The impedances of the microelectrodes in the frequency range of 10 Hz to 20 KHz were measured in saline solution, before and after the electrodeposition of platinum black. Fig. 4 clearly shows that the impedances value (dotted circle) of platinized electrodes at 1 KHz dropped from 8.2 KHz to 2.7 KHz after electroplating of platinum black. Furthermore, all electrodes exhibited a decrease in impedance with increasing frequency, which is a characteristic behavior for a general electrode–electrolyte interface. Scanning electron microscope (SEM) images of the metal surfaces were taken to characterize the performance of our microelectrodes. The electrodeposition of platinum black resulted in a rough surface, thereby increasing its effective area. SEM images of a platinum microelectrode before and after the deposition process are shown in inset of fig. 4, respectively.

C. In vivo Acute Recording

Brain activity was recorded epidurally under two levels of Isoflurane (1.5% and 0.5%), and the spontaneous activity was observed. Fig. 5(c) shows that brain signals recorded with μECoG electrodes are similar to those seen in traditional ECoG electrodes for low levels of anesthesia. The power content of the signal is typical of other surface recording methods (i.e. ECoG or EEG): the peak of the power is seen at approximately 10 Hz (the alpha/mu range), there is a small bump in the power at 20-30 Hz (the beta range), and then decreases monotonically to the high-frequency ranges [17]. Under higher levels of anesthesia, the signal amplitude is greatly reduced, and few if any spontaneous periods of activity are present. The power content of the signal here

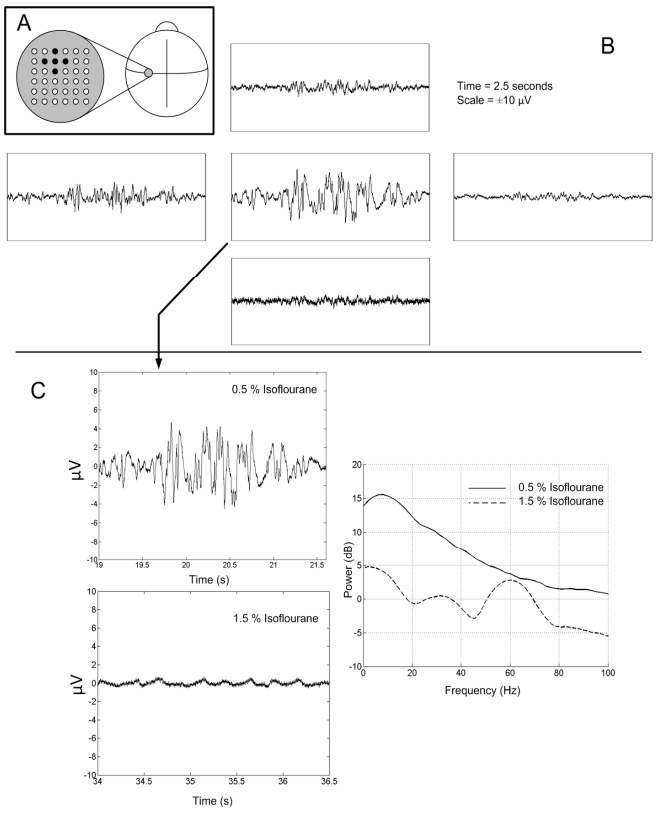


Fig. 5. (a) Location of the burr hole in the cranium, and layout of five selected electrodes. (b) Time plots of the five selected electrodes, recorded epidurally; the time range is 2.5 s, and on a scale of ± 10μV. (c) Larger time plots of the selected center electrode; brain activity under 0.5% Isoflurane (top) and 1.5% Isoflurane (bottom) is compared. Average power spectra for each condition are shown on the right.

shows a greatly reduced alpha-band power (although the peak is still present), and a small peak at the beta-band range is present as well. The 60 Hz noise is present in both recordings, but dominates under high levels of anesthesia due to reduced overall power.

Fig. 5(b) shows recordings from an electrode with a high amount of spontaneous activity, and the signals from the four adjacent electrodes. The center electrode has a signal amplitude of roughly ±7 μV, while the neighboring electrodes 1 mm away show similar activity, but on a smaller scale (±3 μV), or no similar activity at all. This demonstrates that even with inter-electrode distances of 1 mm or less, independent signals can be recorded from a single electrode or small group of electrodes, and that a higher spatial resolution is attainable with the μECoG device than conventional ECoG device.

Generally it was believed that epidural placement on the brain surface is less invasive, and leads to reduced surgical and long-term complications. Subdural placement may provide an increase in spatial resolution and signal amplitude, but may also lead to an increased incidence of infection and other complications, potentially reducing long-term stability. Without further modifications, μECoG electrode arrays designed for cortical recording can be placed subdurally or epidurally on the brain. For the exact comparison between epidural and subdural recording, complementary experiments in which the dura mater is left intact for epidural recording, and then incised for subdural recording will be performed in the future.

Finally, this prototype μECoG electrodes device needs to be verified chemically and mechanically by various evaluation methods. Accelerated soak tests and theoretical calculation of induced mechanical forces may help assess the stability and ensure the safety of the μECoG electrode arrays for long-term implantation.

IV. Conclusion

A μECoG electrode device with a housing platform was designed and successfully fabricated using rapid prototyping and microfabrication techniques. Because of the unique properties of polyimide, this device is flexible and foldable so as to enable epidural implantation through a small burr hole. The size of the implant is significantly smaller than general ECoG grids, and does not require penetrating the brain, as is the case with more invasive microelectrode arrays. Acute recordings from the device demonstrated that independent brain activity could be recorded from electrodes with a spatial resolution of 1 mm. Furthermore, under two different levels of anesthesia, the overall power content of the signal was reduced in higher level of anesthesia compared to lighter anesthesia. These preliminary results suggest that μECoG electrodes may be well-suited for neuroprosthetic applications.

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