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A novel flex-rigid and soft-release ECoG array

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***Abstract*— This article addresses a novel fabrication process for an electrocorticogram (ECoG) electrode array. It consists of three regions: a flexible recording area, a flexible cable, and a rigid field for soldering the connectors. The flexible components can adapt to the curved shape of the cerebral cortex. Furthermore, the entire structure is a free-standing membrane, attached by removable polyimide straps to its carrier substrate. This configuration allows for a high level of control during soldering, electrode characterization and sterilization, as well as a soft release of the array off its carrier just before implantation. The array contains 128 gold electrodes, each 300 nm thick, sandwiched between two 5 µm thick polyimide films. The measuring area of the device is a regular hexagon with a side length of 7.2 mm, designed for implantation on the primary visual cortex of a Rhesus monkey. The flexible cable is 4 cm long. The rigid soldering area was designed for 4×32 OMNETICS connectors. The line resistance from an electrode site to the corresponding electrical connector pin is 540 Ω.**

I. Introduction

Two objectives of modern neuroscience are to understand how networks of neurons produce complex brain functions such as cognition and consciousness, and to apply this knowledge to the clinical and engineering context. This requires high quality interfaces as a main component of neuronal prostheses. They aim at diagnostics and treatment of various neurological impairments, from deafness to epilepsy [1]. Recent results suggest a variety of near-future applications: treatment of non-congenital blindness [2], restoration of movement in paralyzed individuals [3] and enabling disabled patients to communicate with others [4], [5].

Electroencephalography (EEG) can provide control signals for some of these applications; it is easy to use, portable and safe to the patient [4], [6]. Due to its noninvasive nature, EEG is restricted in the spatial and temporal domain. On the other hand, a high spatial resolution is offered by invasive micro electrode arrays inserted into cortical tissue, as they can record the activity of single neurons (spiking). Nevertheless, their main challenges are foreign body responses and safety issues [7]. An electrocorticogram (ECoG) is a good compromise between EEG and spike recording. It uses flat electrode grid arrays implanted either on top or under the dura mater [8], [9]. This provides a higher spatial resolution and less vulnerability to noise artifacts compared to EEG. Moreover, ECoGs have a better long-term stability than single-neuron recordings.

Medical commercial ECoG arrays are relatively large flexible devices with low electrode density e.g. AD-TECH[1]. More advanced high-density arrays consist of thin flexible foils [10]-[13]. Soldering high-density connectors onto such substrates becomes a challenge because of the difficult handling of flexible foils. A possibe solution is to connect the flexible foil to a rigid component as presented in [14] and [15]. However, by introducing an additional (rigid) part the series resistance is increased, as well as the fabrication complexity.

A novel process for fabricating a flexible ECoG array with built-in rigid areas is introduced in this paper. A gold-polyimide array is fabricated on a silicon wafer, whereas the silicon is etched from the backside, everywhere except for the soldering site. Thus, the device can adopt the curved shape of the brain and soldering of electrical connectors can be performed in a similar fashion to fully rigid devices.

II. Design

The principle of operation of an EGoC array is presented in Fig. 1. Electrodes (A) embedded in a flexible material (B) are placed onto the cerebral cortex (C). Brain signals, retrieved by the electrodes, run along the flexible cable through an opening in the skull (D) and to a soldered electrical connector (E). External electronics (not shown), evaluate the brain signals.

In general, ECoG arrays are fabricated as fully flexible devices. They are either peeled off from their carriers or released by etching away a sacrificial film, lying under the arrays. As mentioned before, high-density flexible electrode devices are difficult to handle manually during soldering,

[1] www.adtechmedical.com

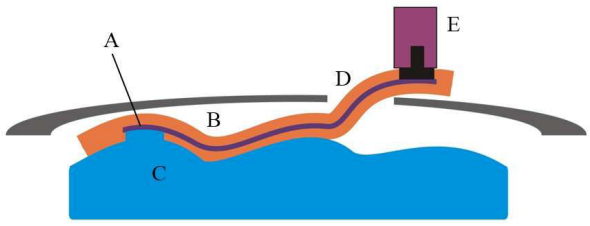

Fig. 1. An implanted ECoG array: electrodes (A), a flexible material (B), brain cortex (C), a skull opening (D) and a soldered connector (E).

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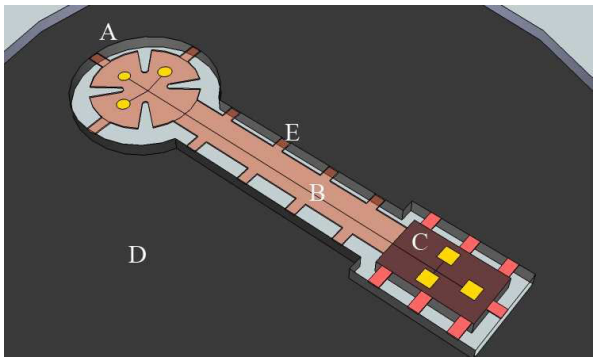

Fig. 2. Array design: a clover-shaped measuring area (A), a cable (B) and a rigid soldering area (C). The array is connected to the Si wafer (D) by polyimide straps (E).

electrical characterization, and sterilization. A possible solution to this problem was reported for a different medical application by Mimoun et al. [16]. It presented rigid sensors connected by flexible interconnects so that the whole structure can be easily folded around and into a catheter.

This paper introduces a novel fabrication process for high-density ECoG arrays which is based on [16]. The design of the array is shown in Fig. 2. It is a free-standing polyimide (PI) membrane, attached to its carrier wafer by several PI straps. By cutting the straps, the device can be released from its wafer.

The layout of the measuring array area is presented in Fig. 3. It contains 124 circular electrode sites and a large reference electrode. The sites have three distinct diameters: 100, 300 or 500 µm. These values are within the range used for other ECoG arrays [13] and are intended to test the effect of different electrode sizes. Most electrode sites are situated on the edges of six concentric hexagons. The spacing between the hexagons is varied—800 µm, 1200 µm and 1600 µm—between the two innermost rings, the second to fourth rings, and the three outermost rings, respectively. This results in inter-electrode distances from 800 to 2200 µm, corresponding to typical values in high-density ECoG arrays [10], [11], [13]. The spatial arrangement of electrodes allows grouping of equidistant electrodes for spatial spectrum analysis. The hexagonal outline of the design ensures symmetry and conforms to the intended region of implantation.

In addition, there are two separate groups of electrodes situated to the left and to the right of the outermost hexagon (Fig. 3). Their electrode diameter is 100 µm and the inter-electrode distance is 400 µm. They are included for high-resolution spatial frequency tests.

The large reference electrode is partially separated from the rest of the device. Thus it can be bent and fixed to the backside of the array, with the metal side facing towards the skull.

The large holes on the reference electrode (inset in Fig. 3) should reduce parasitic capacitance effects induced by the placement of the reference electrode on the array backside. The smaller holes increase the flexibility of the reference electrode. In case of induced mechanical stress, cracks can be prevented from propagating.

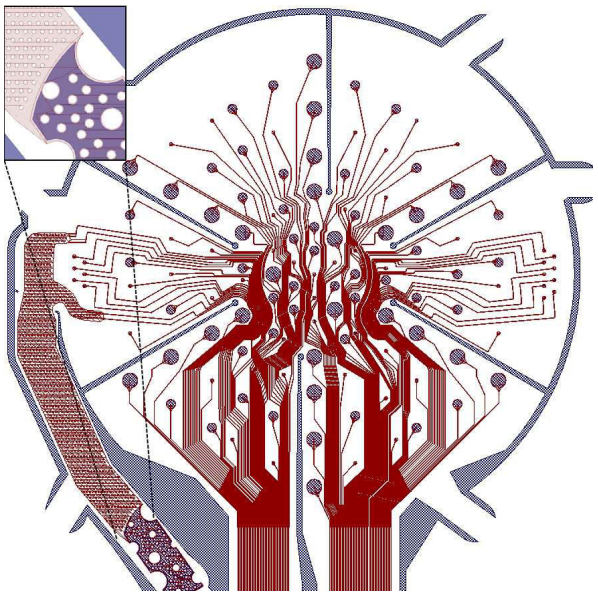

Fig. 3. Layout of the electrode array: electrodes of three different sizes are arranged into six concentric regular hexagons with a step-wise increase in distance between adjacent rings. The inset in the upper left image corner shows a magnification of the reference electrode. The blue regions define openings in the flexible material.

All electrode sites and the reference electrode are connected by metal interconnects to the soldering pads. The minimum center-to-center distance between the lines is 30 µm.

III. Fabrication

The environment of the brain requires flexible and biocompatible materials. Polyimide [10], silicone [11], parylene [12], and silk fibroin [13] have been used for ECoG arrays. Parylene has proven to be a very suitable material due to its minimal uptake of water and its inertness. However, parylene layers are not as robust as polyimide layers of comparable thickness [17]. Silicone is soft and stretchable, and requires low-temperature processing. Polyimide is a compromise between both and it is thus used in this work.

The fabrication of the ECoG array is presented in Fig. 4. An oxidized silicon wafer (380 µm thick Si with 790 nm thick thermal oxide) was used as a substrate. The backside oxide was structured by a conventional photolithography, followed by CF_4 plasma etch. Then a PI film (U-Varnish S[2]) was spun and cured (in N_2, 10 min at 450 °C, 5 µm end thickness). To promote good adhesion, the PI was heat- and plasma-treated (short O_2 plasma) just before the sputtering of a 300 nm thick Au layer. Next the gold was structured and another PI layer was applied and cured. The PI layer was structured and then etched in CF_4 + O_2 plasma in two steps. During the first step the outlines of the array and the straps were defined by pre-etching them to 5 µm PI depth. Afterwards, the PI was structured with another mask and etched again. This time the electrode sites were etched free from the PI and the outlines

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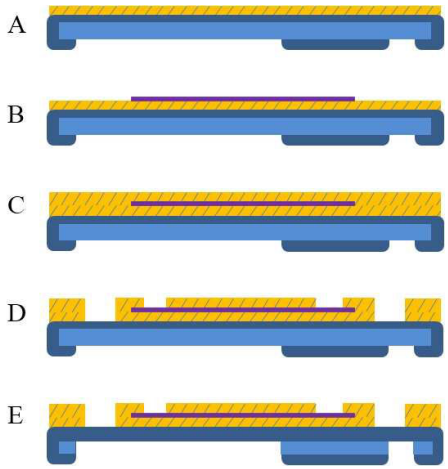

Fig. 4. Array fabrication: A Si/SiO2 wafer is spun with a PI film (A). Gold is deposited and structured (B). A second PI layer is spun (C) and structured (D). The silicon is DRIE etched (E) and the oxide is wet etched.

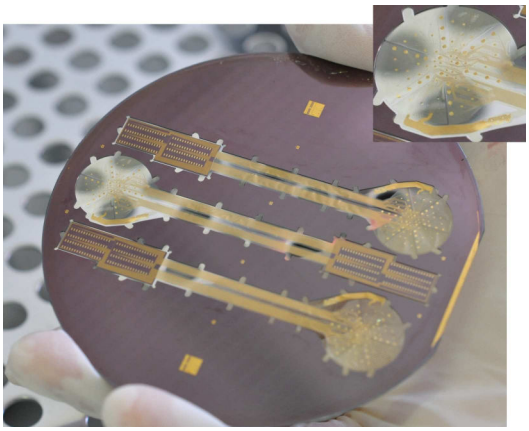

Fig. 5. The fabricated ECoG array: three free-standing polyimide foils are connected to the carrier wafer by polyimide straps. The top-right inset shows the measuring area of the array with the electrode sites situated on concentric hexagons together with the reference electrode.

Fig. 6. The released ECoG array ready for implantation. Its reference electrode is flipped and glued to the array backside. 4 OMNETICS connectors are glued to the rigid soldering area and they are surrounded by an insulating glue.

were completely defined. Splitting the structuring of the PI bilayer into two steps was performed to prevent underetching of the electrode openings. Afterwards, the underlying Si was selectively removed in two steps by a SF_6 + C_4F_8 DRIE process (STS[3] ICP tool with electrostatic clamping). The first etch was 1 h 20 min long. During the second step a Si wafer was glued to the front side with a conductive paste. The second wafer served as a mechanical stabilization for wafer handling. Moreover, it worked together with the paste as a heat sink for the process wafer. Then the stack was etched for another 40 min. The handling wafer was eventually removed and the front-side oxide was chemically wet etched. The fabricated array is presented in Fig. 5.

IV. Soldering and Characterization

First, a conductive adhesive (Elecolit[4] 3653) was manually dispensed on the gold pads, situated on the rigid soldering area. Then, a 32 pin OMNETICS[5] nano connector was aligned and placed on these pads by a FINEPLACER[6] tool. The soldering region was heat-treated for 10 min at 120 °C. Afterwards, for higher mechanical stability, the connector was surrounded by a biocompatible, insulating adhesive (DELO[7] GE-680). Finally, the adhesive was hardened in UV light for 15 min. The last step was repeated after cutting the array out of the wafer (see Fig. 6). The reference electrode was bent and glued to the array backside using a biocompatible adhesive (Polytec[8] EP 653).

The line resistance between the electrode site and the corresponding connector pin was measured before soldering. Its value was 440 Ω. After gluing the electrical connector, the resistance changed to 540 Ω. The parasitic capacitance was measured to be negligible. After sterilizing the array in an autoclave, the resistance altered to 520 Ω.

The array was implanted in the primary visual cortex of a male Rhesus monkey (*Macaca mulatta*). All surgical and experimental procedures were performed in accordance with the corresponding European Communities Council Directive and the regulations for the welfare of experimental animals issued by the Federal Government of Germany, and had been approved by the local authorities. The monkey was trained to perform a fixation task. Fig. 7(a) shows a sample of 2 s of LFP activity, as recorded from eight channels.

V. Conclusion and Outlook

A flex-rigid ECoG electrode array was presented in this article. The novel design and fabrication process offers the following advantages: (1) safer and easier handling; (2) standard soldering techniques can be applied due to the rigid silicon base underneath; (3) direct manipulation of the array is not necessary prior to implantation.

Particularly, the focus will be set on comparing the spatial resolution of epidural and subdural signals, as well as finding the optimal spacing and size of electrodes for field potential recordings.

Once the optimal electrode area and spacing are decided upon, the ultimate goal is to fabricate a “second-generation” array. This array is intended to become part of a fully wireless chronic neural recording microelectronic system.

[3] www.spp-pts.com
[4] www.panacol.de
[5] www.omnetics.com
[6] www.finetech.de
[7] www.delo.de
[8] www.polytec-pt.de

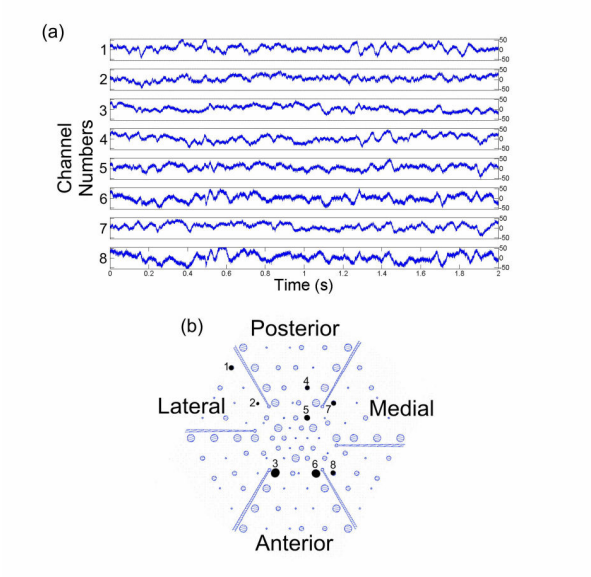

Fig. 7. (a) Sample of 2 s of local field potential activity observed in eight of the 128 channels of the ECoG-electrode array during a fixation task. (b) Layout of the array; numbered black dots show the electrodes associated with the signals displayed in (a).

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