Design and fabrication of all-polymer transducers with different functional features for basic neuroscience and neuroprosthetics

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Abstract

Neural prostheses aim at recording or altering nervous system activity to partly restore motor, sensory or cognitive modalities that have been lost because of disease or trauma. Where appropriate, microelectrode arrays (MEAs) facilitate high-density recordings and local stimulation of neural activity in the brain. Their body integration, functionality and long-term stability may be improved by resorting to new, more tissue-like materials and conductors with low interface impedance and large charge transfer capacity. Recently, we presented an in vitro prototype of a highly flexible polymeric MEA made of a polydimethylsiloxane (PDMS) scaffold with microchannel tracks and electrodes which were coated with films of organic conductors or filled with a graphite-PDMS composite paste [1]. Here, we present an exemplary design concept for in vivo probes with carbon-PDMS conductors based on the same replica-molding technology. They were fabricated from laser-printed templates and feature a particular squeeze-clamping interconnection scheme based on rubber-like contact pads. This “soft contact” strategy alleviates stress-related twist and break found in classically bonded pads in ribbon cable-type wiring to external electronics.

1 Introduction

Microelectrode arrays (MEAs) are among the most commonly used neural interfacing technologies for the recording and stimulation of neural activity. They help in clinical therapy and basic neuroscience research [2]. Traditionally, MEAs are made of materials like glass, metals, or silicon [3]. However, the mechanical mismatch between these rigid substances and the brain tissue is disadvantageous, particularly in scenarios where devices may get displaced as a result of body movement [4]. PDMS is a commonly used flexible carrier that can alleviate this issue to some extent [5]. Despite major advances in neuroprosthetic device technology in recent years, in most cases, material-related failure limits the functional lifetime of an implant. In this study, we designed, fabricated and characterized a tough PDMS-based MEA prototype made from a simple laser-printed template. Its geometry is suited for the recording from within the medial longitudinal fissure in the prefrontal cortex (PFC) of mice.

2 Materials and Methods

2.1 Array design

The MEA µ-channels defining electrodes, tracks and contact pads were sketched with a MEMs CAD design software (Expert, Silvaco) and printed inversely on a laser transparency. They were then molded as 100 µm high µ-topographies using UV-curable nail polish, which was squeezed between a stack made of microscopy slides, the feature- and a black counter transparency and a 100 µm spacer transparency in between. The sandwich was exposed 2 min to UVA and developed for 5-30 min in 3-methylbutanol after removal of both the back-microscopy slide and counter transparency. The resulting µ-topography was used as a molding master for MEA production (Fig. 1a). An array consisted of 12 circular recording sites each with a radius of 80 µm and 310 µm vertical pitch. They were distributed over two square areas. Device dimensions are given in Table 1.

2.2 Material selection and probe assembly

PDMS was chosen as the µ-channel insulator scaffold substrate because of its flexibility, good dielectric characteristics and low tissue response [5]. Carbon-filled PDMS (c-PDMS) was used as a flexible, readily available and easily processable electrical conductor to fill all scaffold voids. The c-PDMS composite was made by adding carbon powder to non-cured PDMS until the DC resistance dropped below 10 kΩ over the distance of about 1 cm. After
spreading c-PDMS into the voids, curing and selectively insulating wires with a thin PDMS backside coat, probes were folded along the shaft edge to slip their pads between standard double row connector pins. Pins had been dipped into c-PDMS to increase and stabilize pad-pin contact. Pad-pins were insulated by an additional coat of PDMS. To increase mechanical brain-insertion stiffness, probe shafts were coated with a thin gelatin film, which was solid at room temperature and slowly dissolved when in contact with water. Once the probe is inserted into the brain, the gelatin coat is expected to dissolve within minutes to hours and may furthermore act as a cell adhesion mediator.

### 2.3 Probe characterization

A perfect recording electrode would feature maximum selectivity and low impedance. The electrical performance of probe electrodes was evaluated and compared by impedance spectroscopy between 1 Hz and 100 kHz in saturated KCl (Perstat 2273 potentiostat, Princeton Applied Research, USA). To ensure that the carbon-filled PDMS was non-toxic to neurons, its biocompatibility was assessed by comparing the health of rat cortex neurons cultured on autoclaved and poly-D-lysine/laminin-coated c-PDMS control MEAs with that of neurons on equally treated, autoclaved and poly-D-lysine/laminin-coated c-PDMS control MEAs with that of neurons on equally treated glass substrates following standard cell culture protocols.

### 3 Results

Absence of cytotoxicity was confirmed with cortex networks up to 14 days in vitro (DIV). In addition, neurons distributed homogeneously on flexible control MEAs without aggregation. Furthermore, cultures survived as long as control cultures on commercial MEAs. As the quality of neural signal recordings is affected by electrode impedance, impedance magnitude and phase were measured. All 8 microelectrodes of one out of three similar prototype probes were functional as exemplarily illustrated in Fig. 2 & 3.

### 4 Conclusions

By using replica-molding microfabrication techniques, we designed and developed a prototype of a very flexible microelectrode array with 8 recording sites that can be implanted into the brain. We successfully tested a particular squeeze-connection strategy for establishing solderless pad-pin contact between the probe and a standard cable connector. To prevent buckling upon probe insertion into the brain, the flexible recording shaft was coated with gelatin to provide temporary insertion rigidity without need for any other delivery vehicle [6]. Impedance spectra indicate that the in vivo probe has a sufficiently low impedance of less than 1 MΩ at 1 kHz and mixed resistive and capacitive properties to allow for recording and electrical stimulation. The electrode arrays will be tested in vivo for their performance in chronic recording studies.

### Acknowledgement

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### References


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**Table 1** Geometries of an in vivo MEA probe

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
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</thead>
<tbody>
<tr>
<td>Total number of recording sites</td>
<td>12</td>
</tr>
<tr>
<td>Number of connected recording sites</td>
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<tr>
<td>Electrode diameter (µm)</td>
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<tr>
<td>Electrode pitch (vertical, horizontal) (µm)</td>
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<tr>
<td>Shaft length (mm)</td>
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<td>Maximum shaft width (mm)</td>
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<td>Wire width (µm)</td>
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