Micromachined Electrode Arrays with Form-Fitting Profile for Auditory Nerve Prostheses

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Abstract—This paper reports the design, fabrication and simulation of a novel micromachined electrode array with a form-fitting profile for use in auditory nerve prostheses. A 10×10 electrode array is created in a 1mm² area using bulk micromachining technology. The space between the individual electrodes within the array is filled with a layer of SU-8 molded to conform to the curved surface of the auditory nerve. This layer enables the implant to be secured to the auditory nerve and to have a good sealing between the array and the nerve tissue after insertion. An electrical model for a single electrode is built. Both mechanical and electrochemical finite element analyses (FEA) of the array are also performed.

Keywords—Electrode arrays; Auditory nerve; Prostheses

I. INTRODUCTION

Direct stimulation of the auditory nerve offers significant advantages over cochlear implants by providing increased spectral resolution and lower power consumption. Human auditory prostheses can be further improved by increasing the number of functional channels and the selectivity and dynamic range of each stimulating electrode. A more accurate tonotopic representation may be functionally restored if an electrode array with very small contact area directly interacts with the auditory nerve, as opposed to prostheses implanted in the scala tympani. However, early attempts in developing intraneural electrodes reported in the 1970s were based on platinum-iridium wire electrodes, which led to insertion trauma and reduced placement accuracy [1].

The development of MEMS technology has made it possible to replace bulky off-chip components with microfabricated counterparts. Microelectrode arrays intended for implantation in the central and peripheral nervous systems have previously been reported by multiple research groups [2-5]. To date, the minimum published shank-to-shank distance has been 400 µm, which is too large for implantation in the 1.5 mm diameter of the auditory nerve near the cochlea.

In our previous work, we reported our efforts to create a high-aspect ratio and high-density electrode array for auditory nerve prostheses [6]. A 10×10 electrode array is created in a 1mm² area using a combination of deep reactive ion etch (DRIE) and HNA wet etch bulk machining technology (Fig. 1).

![Fig. 1 SEM pictures of micromachined high-density electrode arrays.](image1)

In this paper we present the design and fabrication of a novel high density penetrating microelectrode array with a form-fitting profile for auditory nerve stimulation and recording (Fig. 2). A layer of SU-8 polymer is filled into the space between the individual electrodes within the array and shaped using a polydimethylsiloxane (PDMS) mold to complement the surface curvature of the auditory nerve. This layer enables the implant to be secured around the auditory nerve and to physically stabilize it against displacement after insertion. We built an electrical model for a single electrode and analyzed the electrochemical interface design. In addition, we report the results of mechanical and electrochemical FEA simulations of the array, which were performed using ANSYS© software.

![Fig. 2 Conceptual model of the electrode array integrated with on-chip circuitry.](image2)

II. MECHANICAL DESIGN AND SIMULATION

As shown in Fig. 2, the electrode array is intended for implantation on top of the auditory nerve fiber. Due to the curved surface of the fiber, space is left between the tissue...
and the implanted device, resulting in poor sealing. Over a
time, protein fouling will fill the open space, and at a certain
amount, change the position of the implant array. So a good
seal is very important to electrodes for signal recording and
chronic implantation.

We fabricated a form-fitting polymer layer on the
previously fabricated electrode array [6]. After the insertion,
the layer will conform to the auditory nerve surface and
protect against both in-plane and out-of-plane displacements.
Because the molded SU-8 layer progressively decreases the
effective length of the electrodes further from the center of
the array, maximum resistance to bending and buckling is
established for the outermost electrodes, which incidentally
are subjected to the most off-axis loading due to tangential
interactions with the surface of the nerve during implantation.

Mechanical FEA simulation with ANSYS® illustrates the
improvement. Here we consider an electrode array
with/without the polymer curved layer inserted into a volume
of nerve tissue (Fig. 3). We assume that the array is
completely implanted in the nerve fiber, and all deformations
of the tissue are introduced by other movements after surgery,
such as the normal motion of the test subject. In general,
tissue modeling is complex because of inhomogeneous, non-
linear, anisotropic elastic and viscous behavior. In this study,
we assume homogeneous, linear elastic static models that
predict tissue deformations in two dimensions. Such models
are characterized by Young’s modulus and Poisson ratio.
These two parameters are typically determined by testing
small homogeneous tissue samples using rheometers and
similar material testing equipment [7]. The Young’s
modulus and Poisson ratio of the nerve tissue are 34kPa and
0.34, respectively.

10 µN both in-plane and out-of-plane loads were applied
to the electrode in the tissue deformation simulations (Fig. 4).
The simulation data show that, for a given load, the presence
of the curved polymer layer reduced in-plane and out-of-
plane electrode displacements by 10.7% and 9.0%,
respectively, which shows that the polymer form-fitting
profile improves the physical stability of the electrode array
and provides a better seal.

III. FABRICATION

A 10×10 electrode array was created in a 1mm² area
using a bulk machining process (Fig. 5) [6]. The complete
fabrication process will begin with bump bonding a silicon
wafer to a signal-processing and wireless-communication
CMOS chip. This step is not reported here. In this work, we
demonstrated creating 750 µm tall pillars by bulk
micromachining the silicon wafer using DRIE. The pillars
are sharpened into a needle shape with an HNA isotropic wet
etch process. The tip angle of each electrode is less than 20°,
which minimizes the trauma induced during the surgical
operation. The passive array is activated by depositing a
layer of iridium on the electrodes and conformally coating
the tips with a layer of biocompatible Parylene C. The tips
are exposed in a final step by selectively removing the
Parylene C from the tip area.
corsslinked by exposure to UV light (Fig. 6b). The glass structure is removed, and a layer of PDMS prepolymer is poured on the SU-8 replica (Fig. 6d). The PDMS embossing layer is then cured at 70 °C for 2 hours and peeled off the SU-8 mold (Fig. 6e). SEM pictures of the SU-8 replica and PDMS embossing tool are shown in Fig. 7.

![Fig. 6 Embossing tool fabrication.](image)

The process of molding the form-fitting contour is shown in Fig. 8. A 300 µm thick layer of SU-8 2100 was spun onto an electrode array sample (Fig. 8b). The PDMS embossing tool was put above the sample and pressed into the array with an external 400 kPa pressure (Fig. 8c). Then the whole structure was crosslinked by exposure to UV light for 3 minutes. Lastly, the PDMS mold was peeled from the electrode array. The SEM pictures of the electrode array with form-fitting profile are illustrated in Fig. 9.

![Fig. 9 SEM pictures of the electrode array with form-fitting profile.](image)

IV. ELECTROCHEMICAL ANALYSIS

In general, all electrodes for neuron recording and stimulation consist of a metal-electrolyte interface. Compare to the simple ohmic metal-metal contact, a metal-electrolyte contact is a more complex system [8]. An electrode surface can be modeled with a capacitance $C_e$ and resistance $R_e$ of the metal-electrolyte interface in parallel, and a potential $V_e$ dropped across the interface in series (Fig. 10). The value of each component depends on the frequency, material, electrolyte and temperature.

![Fig. 10 A simplified electrical model for a metal-electrolyte interface.](image)

In our project, the electrode is fabricated by highly doped silicon substrate, and then coated with platinum film, which forms the metal-electrolyte interface. A layer of Parylene C covers the whole array except the tips of electrodes to serve...
as an insulation layer. Each single electrode will be connected to the on-chip circuitry with a metal bumper. Based on the model in Fig. 10, the modified electrochemical model for a single electrode with the metal bumper is shown in Fig. 11. Here $R_n$, $C_n$, $R_e$, $C_e$, $R_b$, and $C_b$ are the silicon electrode resistance, insulation capacitance, tip metal-electrolyte interface resistance and capacitance, and bonding bumper resistance and capacitance, respectively. With the similar analyses in [9], all the values can be estimated at low frequency (10 Hz). Compared with the impedance of the metal-electrolyte interface ($R_e$ and $C_e$), the silicon electrode resistance $R_n$ is negligible, while the impedance of the bumper is decided by the standard flip-chip bonding process. $R_n$, $C_n$ and $C_e$ are the main factors to control the electrical model. In order to maximize the signal-to-noise ratio and signal amplitude, we need to minimize the impedance of the interface. We need either to increase the effective area of the electrode tip or to choose a material with high specific capacitance and low charge transfer resistance. Meanwhile, we need minimize the parasitic insulation capacitance by lowering the insulation surface area and increasing the thickness of the Parylene C film. FEA simulations with ANSYS were employed to support the electrical design (Fig. 12).

![Fig. 11 Electrical model for a single electrode.](image)

**V. CONCLUSIONS**

In this study, we report the design, fabrication and of a novel high density penetrating microelectrode array with form-fitting profile for auditory nerve stimulation and recording. A layer of SU-8 polymer is filled into the space between the individual electrodes within the array and shaped to a form that complements the curvature of the surface of the auditory nerve with the use of a polydimethylsiloxane (PDMS) mold. Mechanical FEA simulations show that such a profile provide a better physical stability and sealing, which will benefit to the device chronic implant and the signal-to-noise ratio of the neuron recording. We built an electrical model for a single electrode and discussed the electrochemical interface design of the metalized electrode tip.

More efforts are underway to get better curved SU-8 layers. In addition, experimental work to characterize the electrical property of the electrode array is on-going.

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**REFERENCES**


