

Microelectrode Array Fabrication by Electrical Discharge Machining and Chemical Etching

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Abstract—Wire electrical discharge machining (EDM), with a complementary chemical etching process, is explored and assessed as a method for developing microelectrode array assemblies for intracortically recording brain activity. Assembly processes based on these methods are highlighted, and results showing neural activity successfully recorded from the brain of a mouse using an EDM-based device are presented. Several structures relevant to the fabrication of microelectrode arrays are also offered in order to demonstrate the capabilities of EDM.

Index Terms—Brain-machine interface, electrical discharge machining (EDM), intracortical recording, microelectrode array, neural implant.

I. INTRODUCTION

PROGRESS TOWARD creating brain-machine interfaces (BMIs) for use in humans is accelerating [1], [2]. Some brain-related devices are already in medical use. For example, deep brain stimulator implants are used in therapy to relieve the effects of Parkinson's disease by injecting signals in order to influence brain activity. Auditory prostheses, which stimulate the peripheral auditory nervous system (i.e., the cochlea) in order to restore some sense of hearing, are currently approved for human use and have been implanted in thousands of deaf patients [3]. Considerable quality-of-life improvements could potentially be achieved in the development of BMIs for clinical use. For example, a visual prosthesis could potentially restore partial vision to a blind patient by stimulating neurons in the visual cortex using an input BMI, or signals could be recorded from the motor cortex using an output BMI in order to bypass a neural injury and restore some movement to a paralyzed patient. Even a simpler device that would allow a patient to move a cursor on a screen would make a significant impact. Although much work and study is needed before devices that perform both input and

output can be brought to clinical use, simpler unidirectional devices are likely to emerge much sooner.

It is widely accepted that the cerebral cortex provides the easiest access to motor intent and sensory perception [2], and it is therefore an attractive region for interfacing potential future devices for restoring neurological functions lost due to degenerative muscular diseases, stroke, or spinal cord injury. Studies in nonhuman primates have shown that motion intent of the arm can be anticipated by recording the activity of a few (7–30) neurons in the motor cortex (M1) [4]. Although it has been found that individual M1 neurons are only broadly tuned to a particular direction of arm movement [5], the information can be used to predict movement in a more specific direction [1]. Much is still unknown about the generation of different motor behaviors, but it is generally recognized that clinical applications of such BMIs may require the activities of hundreds or thousands of neurons to be simultaneously sampled [1]. Although not yet ready for use in humans, intracortical recording devices are routinely used to conduct neuroscience research in monkeys and other animals. Mechanically, the devices must gather signals from a number of recording sites, typically arranged in a grid pattern. The need to record from an increasing number of neurons has led to the development of microelectrode arrays [2], [3], [6], [7]. Microelectrode arrays allow one to capture neural signals from a collection of recording sites with a preselected spatial distribution. Many designs and manufacturing techniques have been developed [8], [9]. These devices range from individual wires bonded together to complete structures created using integrated circuit fabrication techniques and complex assembly techniques. Two well-known approaches include the development of microelectrode arrays at the University of Utah [10]–[13] and at the University of Michigan [14], [15].

In order for a neuroprosthesis to reach widespread use, it must be manufactured at a large scale, in an automated fashion, and with reliable yields [16]. Electrical discharge machining (EDM) is a process that makes use of computer-aided design (CAD), that runs under computer numerical control (CNC), and that is capable of batch processing. It can generate intricate features with high aspect ratios and is capable of machining a large variety of conductive materials. In this report, the development of EDM for use in the fabrication of microelectrode arrays is explored and assessed.

II. METHODS

A. EDM

Microelectrode arrays were machined using a Charmilles Technologies Robofil 1020 series wire EDM and Charmilles SW-10-A zinc coated brass 100- μm wire. A workpiece was

Manuscript received July 2, 2003; revised January 16, 2004. This work was supported by the NIMA Foundation and in part by: the National Institute of Neurological Disorders and Stroke (NINDS) of the National Institutes of Health (NIH) Contract N01-NS22345, NIH/NINDS under Grant NS-25074, and the Defense Advanced Research Projects Agency (DARPA) under Grant MDA972-00-1-0026. *Asterisk indicates corresponding author.*

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Digital Object Identifier 10.1109/TBME.2004.826679

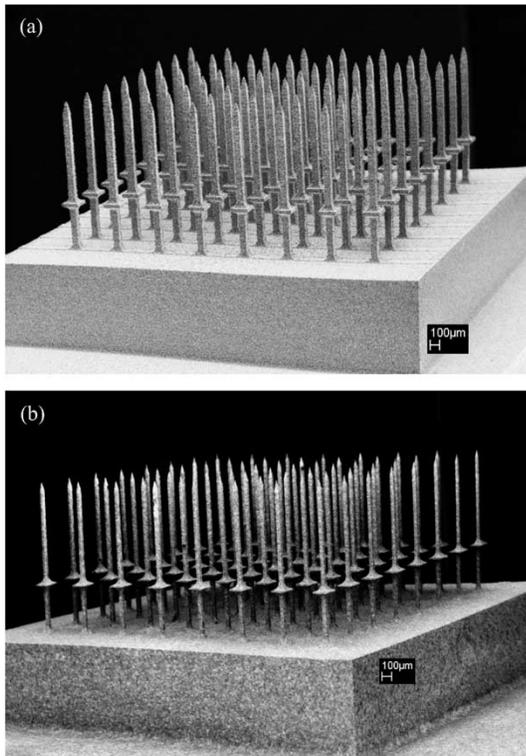


Fig. 1. SEM images of an electrical discharge machined titanium microelectrode array that has undergone the chemical etching process. (a) Before etching. (b) After etching.

mounted in the EDM tank perpendicular to the EDM wire, and an initial cut was performed through one plane [17], [18]. The cut consisted of three passes operating with a pulse duration of $0.2 \mu\text{s}$, machining voltages of -80 V , -80 V , and $+120 \text{ V}$ with respect to the workpiece, and machining frequencies of 200 kHz , 1 MHz , and 1 MHz , respectively. The workpiece was then rotated 90° , and the process was repeated to produce a second cut orthogonal to the first. In this way, electrode arrays have been machined from titanium, titanium-aluminum-vanadium alloy (Ti90-Al6-V4), stainless steels, and tungsten carbide, although the process should be compatible with most conductive materials, including doped silicon.

Fig. 1 shows scanning electron microscope (SEM) images of titanium microelectrode arrays that were fabricated by EDM. Ledges were machined as part of the electrodes in order to aid in later assembly. The electrodes were machined in order to obtain final lengths of 1 mm , initial electrode widths of $80 \mu\text{m}$, and an inter-electrode spacing of about $500 \mu\text{m}$. Fig. 2 shows an SEM image of a more elaborate 10×10 microelectrode array that was similarly fabricated from titanium alloy and was designed to accommodate the mounting of an electronic die [19]. The array itself has an inter-electrode spacing of $400 \mu\text{m}$, but the rows of electrodes at the base of structure, which are later encased in epoxy and ground down to form pads for gold bonding, are spaced $250 \mu\text{m}$ apart. Fig. 3 shows an SEM image of an 1141-electrode honeycomb-shaped array of hexagonal electrodes machined from titanium alloy by performing three EDM cuts and rotating the workpiece by 60° between cuts. Each electrode in this array is $250 \mu\text{m}$ from its six neighbors. Arrays of electrodes

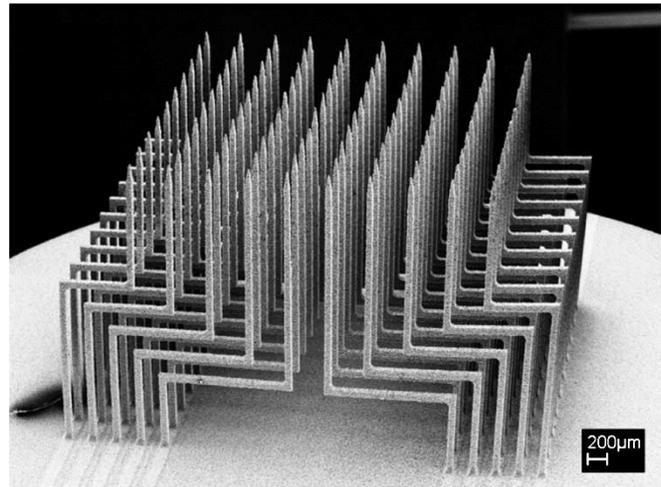


Fig. 2. SEM image of a 10×10 microelectrode array that was electrical discharge machined from titanium alloy. This array, designed to accommodate the mounting of an electronic die, has an inter-electrode spacing of $400 \mu\text{m}$. The rows of electrodes at the base of structure, which are later ground down to form pads for gold bonding, are spaced $250 \mu\text{m}$ apart (from [19]).

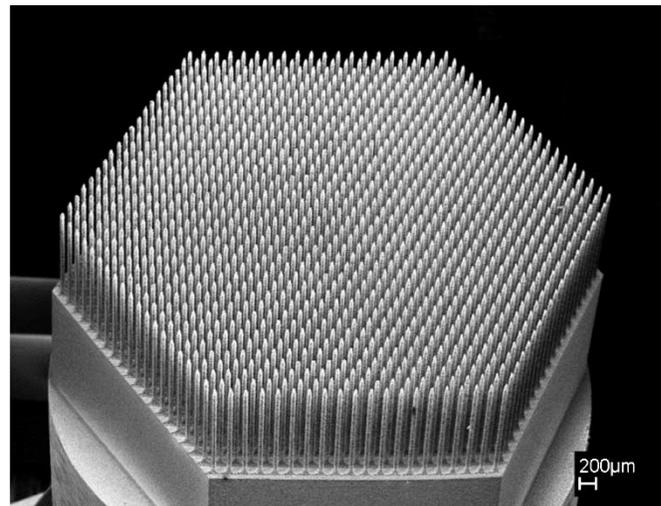


Fig. 3. SEM image of an 1141-electrode titanium alloy microelectrode array that was electrical discharge machined in a honeycomb pattern (from [19]).

that have a range of lengths have been fabricated [18]. Electrodes with lengths exceeding 5 mm and inter-electrode spacings of less than $250 \mu\text{m}$ have also been machined using these methods [18]. The spacing is limited by the size of the wire used in the EDM process. The structures shown in this report were machined using a $100\text{-}\mu\text{m}$ diameter EDM wire, for which the minimum cut size was found to be approximately $170 \mu\text{m}$. Closer inter-electrode spacings could be obtained using a $30 \mu\text{m}$ diameter EDM wire, for example.

B. Chemical Etching

After the microelectrode arrays were machined by EDM, they were electroplated with an acid gold strike, followed by platinum. In order to remove any oxide layer present on the electrodes, the arrays were placed in a hydrochloric acid (HCl , 37% approx.) bath heated close to its boiling point immediately before being placed in the electroplating solution. This acid bath

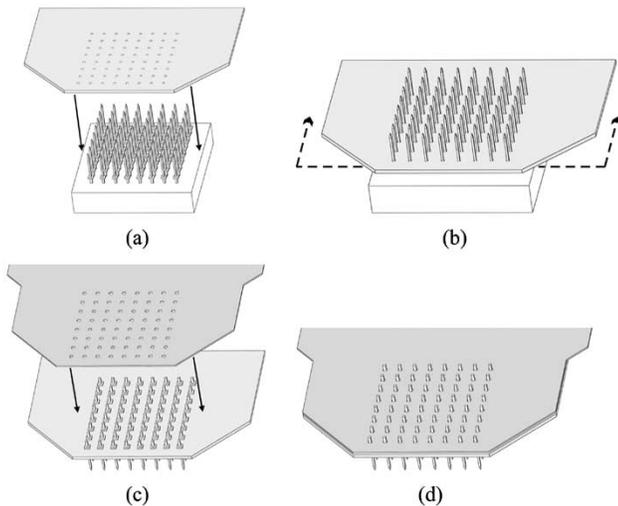


Fig. 4. Schematic showing the steps in the assembly process. (a) A polyimide insulating substrate, coated with a thin layer of epoxy, was lowered onto the microelectrode array until it rested on the ledges of the electrodes. (b) After the epoxy cured, the array was removed from the metal base by wire electrical discharge machining. (c) The electrode assembly was flipped over, and the flexible printed circuit board connector cable was lowered over the back ends of the electrodes. (d) The electrodes were soldered to electrical pads (not shown) that surrounded each of the through-holes in the connector cable. The solder joints were sealed with epoxy and parylene was vapor deposited over the entire array assembly. Parylene was then removed from the electrode tips by laser ablation in order to create the recording sites.

was additionally found useful as an etching process in order to reduce an array's dimensions [18]. In the case of a titanium array, etching was found to occur uniformly at a rate of roughly $2 \mu\text{m}/\text{min}$. Fig. 1 shows SEM images of a titanium microelectrode array that has undergone the chemical etching process. This method was found to be a good complement to EDM. It reduces the need for the later passes in the wire EDM cuts because the final surface finish is dependent on the etching process. It can also further enhance the aspect ratios of the electrodes. For example, by chemically etching an array of 5-mm electrodes to electrode widths of less than $50 \mu\text{m}$, one is able to create electrodes with aspect ratios of greater than 100:1.

C. Device Assembly

Devices were designed and fabricated for use as a chronic implant in a *Macaca Mulatta* monkey [20]. The region of interest for the intended recording is at a depth of 1 mm below the surface of the brain. The array configuration shown in Fig. 1 served as the basis for the implant. Platinum, electroplated over titanium alloy, was chosen as the surface metal for the device due to its characteristics in recording bioelectric activity, as well as its biocompatibility [8]. Fig. 4 shows a schematic of the assembly process used in constructing the devices. An assembled device is shown in Fig. 5.

D. Encapsulation

The electrodes must be electrically insulated in order to obtain desirable electrical characteristics. Parylene C was chosen as the insulator due to its biocompatibility and the ability for it to be vapor deposited uniformly. A $3\text{-}\mu\text{m}$ layer of parylene was used as an encapsulant over the entire microelectrode



Fig. 5. Image of a fully assembled implant.

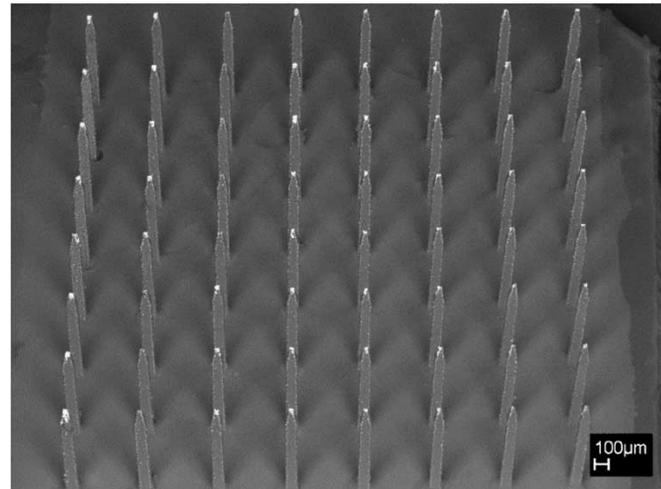


Fig. 6. SEM image of a parylene-coated assembly, consisting of platinum-coated electrodes epoxied into a polyimide substrate, after the electrode tips have undergone laser ablation. The array tips are bright in the images due to the higher electron densities recorded by the SEM detector of the exposed platinum relative to the parylene.

array structure. To expose areas at the tips of the electrodes as recording sites, laser ablation was performed using a Resonetics Maestro 1000 excimer laser operating at a wavelength of 248 nm in order to achieve impedance magnitudes in the $100 \text{ k}\Omega$ to $200 \text{ k}\Omega$ range at 1 kHz when measured in 0.9% saline using a platinum reference electrode. This range of impedance magnitudes has been found suitable for intracortical recording [9] and corresponds to an exposed recording site diameter of about $20 \mu\text{m}$. The laser was operated at a pulse energy of 160 mJ, a pulse rate of 200 Hz, and a pulse width of $1 \mu\text{s}$. An iris was used to control the size of the ablated region and therefore blocked some of the energy. The number of pulses executed per electrode was varied from 20 to 100 in order to generate a range of impedance values. Fig. 6 shows an SEM image of a parylene-coated assembly after the electrode tips have undergone laser ablation.

E. Surgical Procedures

A device identical to that shown in Fig. 5 was implanted into a mouse in order to validate the electrical properties of the design. The mouse was anesthetized, secured, and prepared for the surgery, and a craniotomy was performed on the animal. The microelectrode array was then positioned using a micromanipulator over the left side of the animal's brain and lowered directly downward until the electrodes slowly penetrated into the brain. Because a mouse's dura mater is very thin compared with that of a monkey, a decision was made to make the array penetrate through the dura mater. Because the array covers the greater part of a hemisphere of a mouse's brain, it was expected that only a few electrodes would be at a suitable recording depth at a given insertion depth due to the brain's roundedness. Recordings were made at several depths on both hemispheres during the procedure.

III. FABRICATION RESULTS AND ELECTRICAL CHARACTERIZATION

The EDM-based fabrication steps were found to deliver yields of near 100%. The use of clamps specially designed for the fabrication of the microelectrode arrays ensured that the arrays could be fixtured and removed from the EDM without being damaged. Machining time for a 64-electrode titanium alloy microelectrode array from a 10-mm diameter rod was typically about 3.25 h. By machining multiple arrays simultaneously as a batch process, however, this machining time can be reduced substantially. For example, 49 100-electrode arrays were simultaneously machined from a square pattern of 49 5-mm by 5-mm elevated regions at a rate of about 0.5 h/array.

The yield concerning the range of obtained electrode impedance values is perhaps more important, however. This yield is dependent on many variables, including the uniformity of the parylene deposition and the insulation removal results from laser ablation, as well as the integrity of the solder and wiring connections. Electrical characterization of the electrodes was performed by obtaining impedance curves for each electrode using a Hewlett Packard 4194-A impedance analyzer. Measurements were taken at 25 °C in 0.9% saline using a platinum 30-mm-square reference electrode. The 64 electrode impedance values were found to be bimodally distributed with 43 electrodes having much lower impedance magnitudes at 1 kHz (mean = 78 k Ω , standard deviation = 76 k Ω) and 21 having impedance magnitudes of several M Ω at 1 kHz (mean = 5.7 M Ω , standard deviation = 3.2 M Ω). This distribution of values is believed to be due to poor solder connections, exhibiting a yield of 67%. The use of titanium alloy as the electrode material may have contributed to the poor soldering result by limiting solder adhesion. The group of 43 electrodes had a distribution of impedance magnitudes thought suitable for an *in vivo* validation.

IV. NEURAL RECORDING RESULTS

Some dimpling of the brain was observed upon insertion of the electrodes using the micromanipulator, but it later subsided. No bending of the electrodes was observed during the procedure. Data was collected at several depths using a Cyberki-

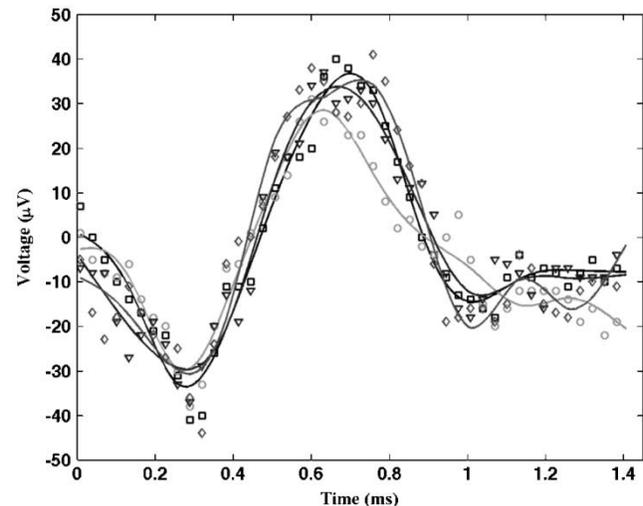


Fig. 7. Plots of neural spikes recorded from the cortex of a mouse.

netics 128-channel Cerebus data acquisition system designed for intracortical recording extracellular action potentials. The system software is capable of filtering, sorting, and recording neural spikes over all 64 electrodes in real time. By setting the thresholds and waveform filters, one could selectively display the data of interest. Neural spikes normally have durations of 1 to 2 ms and amplitudes of about 100 μ V peak-to-peak, and they can fire at rates of up to 300 Hz but often do so much less frequently. Waveforms of this description were observed using the implanted device. Four such waveforms, recorded in the mouse on a single electrode, are shown in Fig. 7. The markers represent data that were recorded at 32 kHz. The signals were low-pass filtered at 7.5 kHz and high-pass filtered at 750 Hz while recording. Interpolated curves are shown for clarity. As expected, due to the distribution of impedance magnitudes and the geometry of the procedure, characteristic neural spikes were observed on only a subset of the electrodes at a given penetration depth into the brain. Characteristic waveforms were observed at a variety of depths, however, and in both hemispheres of the brain during the procedure. Neural activity was found to increase as time elapsed between anesthetic doses. Larger bursts of action potentials were additionally observed on a few electrodes when the animal exhibited slight movement or twitched.

V. SUMMARY AND FUTURE WORK

Newly developed methods of fabricating microelectrode array assemblies have been presented. Electrical discharge machining was found to be a suitable process for creating features with sizes and aspect ratios common to microelectrode array assemblies. The use of new methods such as epoxying a preformed structure onto features machined into the electrodes themselves, and the use of complementary technologies such as chemical etching, parylene coating, and insulation removal by laser ablation have also been demonstrated. Many questions regarding optimum array geometries and materials remain. The flexibility and repeatability gained by EDM as a computer numerically controlled process compatible with today's CAD programs make it an attractive technology for developing new electrode geometries and configurations.

Initial studies have been performed in order to demonstrate the suitability of a device fabricated with the above methods for neural recording. Neural activity was observed and recorded in a mouse in an acute experiment. Although neural spikes were observed on some of the electrode channels, many others exhibited only noise. This is attributed to the use of hand soldering as a method of connection and the use of open-loop control in generating the electrode tips by laser ablating the parylene insulation. More work is required in order to better calibrate and automate the laser ablation process. Many connecting and packaging issues also need to be resolved, including an improved soldering technique or a replacement for conventional soldering such as the use of wire bonding or flip chip technology. Methods of preparing the back ends of the electrodes prior to electrical connection, such as grinding them flat and coating them, are being explored [19]. It is hoped that the methods presented here will develop into a useful tool for biomedical and neuroscience research and help in the development of a future clinical neural interface.

ACKNOWLEDGMENT

The authors would like to acknowledge J. Maláček, J. Burgert, B. Dyer, and C. Wiseman of the Massachusetts Institute of Technology for their roles in this work, M. Serruya and M. Fellows at Brown University for their input, and N. Issa and A. Mallik at the University of Chicago for their help in acquiring the data.

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