DESIGN AND FABRICATION OF A MICROELECTRODE ARRAY DEDICATED FOR CORTICAL ELECTRICAL STIMULATION

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Abstract

A manufacturing process has been developed for a microelectrode array that can be used for stimulation of the visual cortex to provide an effective sense of sight to the blind. The first prototype merges sixteen 316L stainless steel electrodes on a 4x4 array. Several arrays can then be assembled together in a mosaic configuration to cover a large portion of the visual area. To reduce brain injuries while maximizing the quality of the phosphenes, the electrodes must be sharp, thin (50 µm-diameter), separated 400 µm apart and have stimulation sites located at 1.5 mm under the surface of the cortex. Using electrical discharge machining followed by an electrochemical surface treatment, it has been shown that it was possible to reach such geometrical ratios at this scale. This technique can be combined with electrodeposition of porous platinum to create low impedance stimulation sites. Encapsulation and biocompatibility of the array are also discussed.

Keywords: *Microelectrode; micromachining; neural interface; cortical stimulation; visual prosthetics.*

1. INTRODUCTION

The Polystim research team from École Polytechnique de Montréal is presently working on the design of a wireless visual stimulator [1]. This smart medical device could restore sight for blind patients by stimulating directly the visual cortex. Such an approach requires a reliable brain/device interface adapted to an environment that is as harsh as it is delicate.

It has been known for years that, in order to stimulate the cortex to produce well-defined phosphenes (pixels that appear in the visual field) with low currents, the best procedure is to use microelectrodes that penetrate the brain [2]. Such three-dimensional electrodes show many advantages over the first generation of planar electrodes that were floating at the cortex surface. Indeed, thin penetrating electrodes that have stimulation sites reaching a depth of approximately 1.5 mm can produce phosphenes with only a few tens of microamps, which is 10 to 100 times less than for surface electrodes. Hence, the required tension can be diminished significantly, which is essential for a wireless lightweight device. The 2-mm minimal distance between the stimulation sites can also be reduced to approximately 250 µm before superimposing the phosphenes. Image resolution provided by this kind of neural interface will therefore be greatly improved.

Following this pioneer work, a few teams opted for the penetrating electrodes solution and two main designs took shape. First, Wise *et al.* from University of Michigan proposed an approach based on silicon microelectronics technology [3, 4]. Using selective doping and chemical etching, two-dimensional probes are built. Then, the 2-D probes can be assembled together in a three-dimensional array with the use of a platform and spacers. The interesting aspect of this technique is that it offers the possibility to integrate electronics directly on the probes or on the platform. Also, many stimulation sites can be placed on a single electrode. It is then possible to stimulate the cortex at various depths, which enhances the probability to have a site close to a responsive group of neurons. These sites can be placed far from the tip of the electrodes, where brain tissues suffered the most damage. However, the technology used for this implant is complex and requires expensive facilities and great control of the manufacturing process. The assembly of the array is also problematic because micromanipulators must be used and reliable 90° electrical connections between the probes and the platform have to be made.

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A second approach was investigated by Normann *et al.* from University of Utah [5, 6]. Using a precision dicing saw, perpendicular grooves are cut in a doped silicon die and filled with melted glass. A second set of grooves is then cut on the other side of the die to reach the glass layer and electrically insulate the electrodes. Chemical etching, metal deposition at the stimulation sites and at the back of the electrodes and biocompatible encapsulation complete the process. The main advantage of this technique resides in its simplicity. A complete array can be built in less time than a microelectronics-based array and the equipment used is relatively simple, which makes the arrays less expensive to produce. Even though the Utah intracortical array is limited to one stimulation site at the tip of each electrode, the Polystim team believes that a simple reliable device should be privileged. Therefore, we have chosen to build an array inspired by this design.

Stimulating the visual cortex to provide the blind with useful vision is probably the most spectacular application foreseen for brain interfaces. However, considering a few design changes, a visual stimulator array could be used with other neuroprosthesis, such as motor or auditory stimulators or cortical signal recorders. This illustrates the need for a cortical array that can be easily adapted to any of these devices.

2. MANUFACTURING AND RESULTS

2.1 Electrical Discharge Machining

Although wire electrical discharge machining (wire-EDM) has been used for decades, the MEMS industry has discovered its potential only a few years ago when these machines were equipped with durable thin wires and high performance sensors and actuators. The principle of operation of the wire-EDM is to apply a strong potential between a conductive workpiece (the electrode array) and a metallic wire. When the distance between the two is small enough, electrical arching occurs and blasts a thin layer of the workpiece. The expulsed material is then removed with a dielectric fluid and the process is repeated several thousand times a second as the wire penetrates the workpiece. Hence, there is no physical contact between the wire and the array that could damage the electrodes. This represents a great advantage over the dicing saw that exerts a force that can bend or break the electrodes. Thus, only a small number of hard conductive materials can be used for the Utah cortical array and the electrodes can hardly exceed 1.5 mm in length. This method limits the possible applications of the array. Wire-EDM is far less restrictive and can cut high aspect ratio structures effectively in most metals. It is even possible to cut the structures in a hexagonal configuration to improve electrode density. The distance between electrodes is fully adjustable and can be reduced easily to the minimum 250 µm. The main limitation of the EDM process is the dissipated heat due to the electrical arching that can bend electrodes thinner than 80 µm. Our prototype has 4x4 1.7 mm long, 100 µm wide electrodes that are separated 400 µm apart.

Even if EDM can basically cut any conductive material, some are more suitable for achieving a good surface quality. For example, medical grade 316L stainless steel gives good machining results and is also one of the most commonly used metals for traditional cortical electrodes and other types of implant. The following steps of the process also work effectively with this relatively cheap material, as will be discussed in the next sections.

Wire-EDM is a slow process that can take up to 6 hours to create a 4x4 array, depending on the desired surface finish. However, this automated equipment can work on its own from a CAD file, saving time and assuring high reproducibility. It is also possible to machine several arrays at the same time, considering a few extra steps. Fig. 1 presents the two-pass process resulting in a 4x4 array.

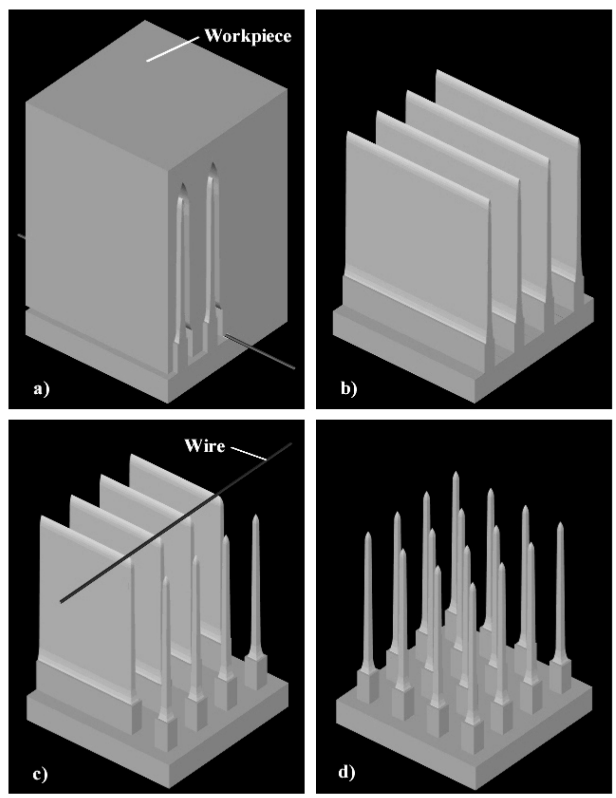

Fig. 1. Array machined by 2-pass EDM process (a) first pass in progress; (b) first pass completed; (c) second pass in progress; (d) second pass completed.

2.2 Electropolishing and Electrodeposition

A rigorous optimization of the parameters in the EDM process allows a surface finish of 0.1 µm Ra on 316L. Unfortunately, it is quite difficult to keep a good control on these parameters. Also, the arching between the wire and the workpiece always produces a thin layer of thermally degraded metal that has a porous aspect, as seen on Fig. 2:

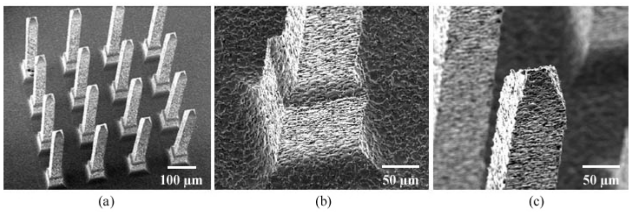

Fig. 2. Thermally degraded layer following electrical discharge machining (a) full view of a 4x4 array; (b) lower part of an electrode; (c) tip of an electrode.

The porosity of this layer makes it an ideal site for pathogen proliferation, which may eventually cause complications. Also, a rough surface might damage biological tissues during insertion and presents stress raisers that weaken the structures. An electrochemical surface treatment in an acid bath is an efficient way to remove this layer. By adjusting the voltage and the treatment time, it is possible to make the oversized electrodes fit the ideal dimensions for implantation (40 to 50 µm wide, 1.5 mm long). One must consider that the electrodes diminish faster in length than in width, since the etch rate is regulated by the current density. Fig. 3 presents the same views as Fig. 2, after electropolishing.

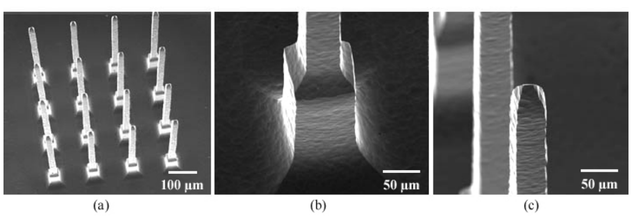

Fig. 3. Thermally degraded layer removed by electropolishing (a) full view of a 4x4 array; (b) lower part of an electrode; (c) tip of an electrode.

Electropolishing prepares the surface adequately for the subsequent electrodeposition of platinum at the tip of the needles. To keep the impedance, and power consumption, as low as possible during operation, the metal must be porous to get a high effective surface. However, platinum is more delicate in its porous form than in its hard, smooth form. *In vivo* tests will then be needed to find a suitable porosity that can resist the insertion and the friction in the cortex over a long period of time. An experimental setup is presently tested for impedance measurements. According to experiments made on similar devices, the impedance is expected to be of the order of 100 kΩ at 1 kHz. Our team is also investigating the possibility to put the stimulation sites away from the tips, where compressed tissues might be less responsive to electrical stimulation.

2.3 Electrical and Biocompatible Insulation

After the wire-EDM step, the electrodes are still electrically linked together. To insulate them, a thin layer (300 µm) of low-viscosity biocompatible epoxy is spread on the base of the electrodes. The low permittivity epoxy is then cured to form a solid platform. Finally, the stainless steel base linking the electrodes is mechanically polished, as seen on Fig. 4.

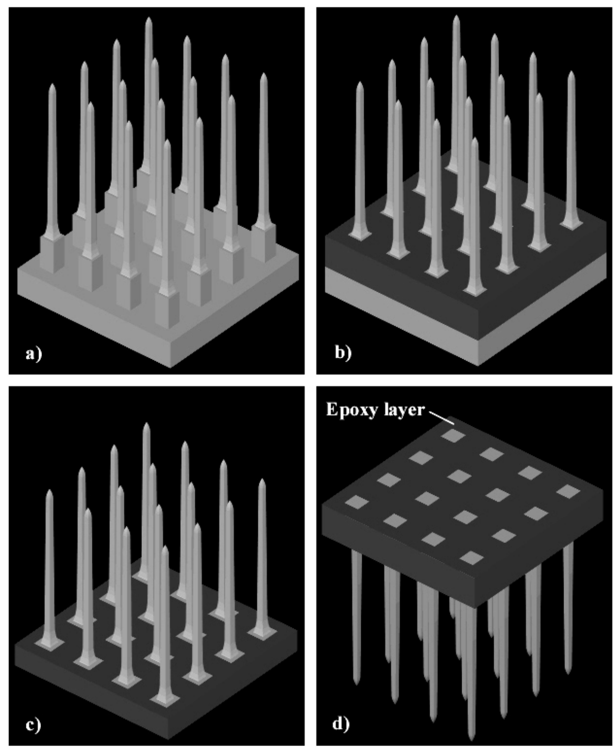

Fig. 4. Electrical insulation of the electrodes (a) electropolished array; (b) layer of epoxy at the base of the electrodes; (c) top view after polishing the metal base; (d) bottom view after polishing the metal base.

The biomedical industry is still looking for the ideal encapsulation material for chronic implants. Presently, Parylene-C is probably the best polymer used for biocompatible insulation, but it still has to prove its suitability for long-term applications. While waiting for new progress in biomedical encapsulation, Polystim decided to use a 2-3 µm thick layer of Parylene-C to cover the whole array, except the stimulation sites and the electrical contacts at the back of the electrodes. Many other teams made this choice, such as University of Utah, Illinois Institute of Technology and Massachusetts Institute of Technology [7, 8].

The particular geometry of a device as small as a cortical array makes it difficult to use conventional techniques of encapsulation like dipping or spraying. Vapor-deposited material processes are usually very directional and are therefore inappropriate for the three-dimensional structure of the array. Parylene deposition uses short mean free path monomers that move in the deposition chamber in an erratic way. Molecules can then reach hidden surfaces and combine with other monomers to constitute a uniform Parylene layer.

This process results in the complete encapsulation of the electrodes, including the stimulation sites. To uncover these, a technique similar to that of University of Utah is used [6]. The electrodes of the Parylene-coated array are pushed through a thin metal foil with a micrometric device, exposing approximately 100 µm of the tips. A metallic box protects the rest of the array and the whole apparatus is inserted in an oxygen plasma chamber. The Parylene is completely etched off the tips, revealing the stimulation sites.

3. CONCLUSIONS

A process for manufacturing an intracortical array has been developed. Although this array is specifically designed for stimulation of the visual cortex, the process parameters are flexible and can vary over a wide range. The array can then be easily adapted to other kinds of brain recording or stimulating devices. Electrical discharge machining of medical grade 316L stainless steel has shown to be a reliable method to produce high aspect ratio structures. These structures can be electrochemically etched to reach the dimensions required to minimize damage to the brain during insertion. Electropolishing also prepares the surface for electrodeposition of platinum by cleaning, leveling and removing the oxide at the surface of the metal. An epoxy platform and a biocompatible layer of Parylene-C that is etched off the stimulation sites complete the fabrication of the array.

Our first prototype still has to be tested to validate the fabrication steps. The dimensions of our array seem to be appropriate for implantation, but the stainless steel electrodes will have to be strong enough to be inserted without bending or breaking. *In vivo* testing will be necessary to find the appropriate platinum coating for the stimulation sites. Porous sites show the best electrical properties, but since they are less resistant, adherence will have to be optimized.

The technique used to connect the array to the rest of the intracortical implant is presently in development. Some possibilities are considered and they might slightly affect the steps described above. As soon as the connection is complete, clinical tests will begin on small mammals like rats and cats to verify the effectiveness of the stimulator.

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