J. Micromech. Microeng. **14** (2004) 104–107 PII: S0960-1317(04)60340-8

3D flexible multichannel neural probe array

Shoji Takeuchi[1], Takafumi Suzuki[2], Kunihiko Mabuchi[2] and Hiroyuki Fujita[1]

[1] Center for International Research on Micro Mechatronics (CIRMM), Institute of Industrial Science (IIS), The University of Tokyo, Tokyo, Japan
[2] Department of Information Physics and Computing, The University of Tokyo, 4-6-1 Komaba, Meguro-ku, Tokyo 153-8505, Japan

E-mail: takeuchi@iis.u-tokyo.ac.jp

Received 4 March 2003, in final form 2 July 2003
Published 14 October 2003
Online at stacks.iop.org/JMM/14/104 (DOI: 10.1088/0960-1317/14/1/014)

Abstract
A 3D flexible multichannel microprobe array was designed, fabricated and tested. Since each probe had several recording pads, the probe array could be used to measure neural activity at various depths in the brain. They were batch fabricated with interconnections, using a specific folding process to fold the planar probe structures. This flexible probe array was inserted into a rat's brain without fracturing and was successfully used to measure neural signals.

1. Introduction

Neural recording electrodes are a typical application of MEMS devices. In the past decade, many types of electrodes have been reported [1–3]. These have mostly used silicon structures for their insertion probes and substrate because silicon can be fabricated to a small scale with multichannel recording sites on tiny probes as well as having enough strength to be implanted into neural organs at precise locations. However, since the material does not deform in the organs, the recording position shifts slightly and consequently damages cells when the body moves. This is why this kind of rigid electrode is not suitable for long-term neural recording.

On the other hand, recent MEMS devices have been able to employ flexible materials, such as polyimide, SU-8, parylene and PDMS. They can also be fabricated in a microsize scale through lithography, micro-molding or dry etching technologies as well as the silicon materials. The advantage of these soft materials is that they can deform shape without fracturing compared to rigid materials such as silicon or silicon dioxide. Using these kinds of materials, flexible electrodes have been reported by various groups and they can attain implantation with less damage [4–6]. In this paper, a flexible probe array type microelectrode is proposed that can be fitted into organs to achieve multichannel neural recording from three-dimensional (3D) positions.

Figure 1 illustrates the concept behind our 3D flexible-probe array. Compared to the traditional probe arrays [1], the advantage of our electrode is not only its flexibility but it can be used to measure brain activity at various depths simultaneously due to several recording pads on each probe. This means that our probe array should be useful in obtaining 3D data on brain activities in a single insertion. It is difficult to implement such recording pads on a probe with conventional bulk-silicon micromachining. Having overcome this problem, Hoogerwerf and Wise reported on the manual assembly of 2D probe arrays [7]. Here, we proposed a batch assembly method using surface micromachining which was a process of bending 2D probes.

2. Design and fabrication

A magnetic field was used for the batch assembly, which has previously been presented by the authors [8], because it avoids complicated manual assembly and saves time. The magnetic assembly in this work starts with the patterning of 2D probes on a plain substrate (figure 2(*a*)). Multiple recording pads and interconnections can be formed on each probe at this stage. These probes have a magnetic thin plate on their backsides. When the external magnetic field is applied to the probes, they stand up towards the direction of the magnetic field. The recording pads of the probes are vertically aligned in this way. In a test experiment on the standing process, the probe was bent to 90° when an external magnetic field of 380 mT was applied (figures 2(*b*) and (*c*)). In this experiment, we used electroplated

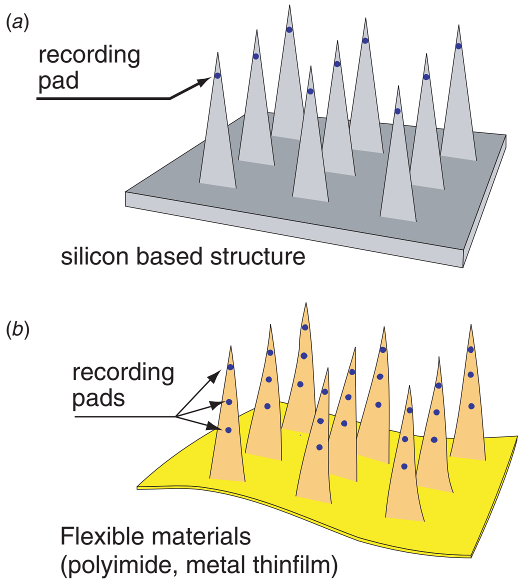

Figure 1. (*a*) The traditional electrode has only a single channel at a probe made of silicon. (*b*) We propose a totally flexible multichannel probe array. Each probe has several pads and can measure the neural activity at various depths of the brain.

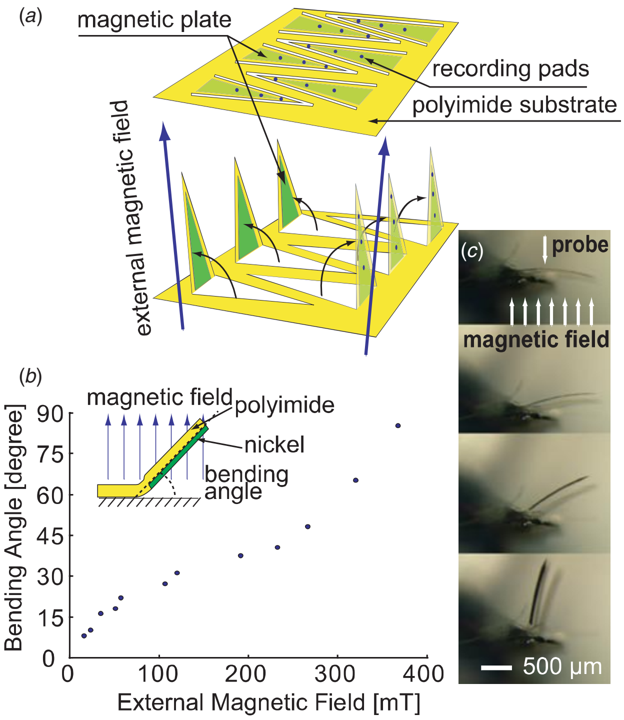

Figure 2. Magnetic batch assembly of the probes. (*a*) The idea of the batch assembly of standing probes. (*b*) Result of a bending experiment of the probe structure using an external magnetic field. (*c*) Photos of the bending experiment.

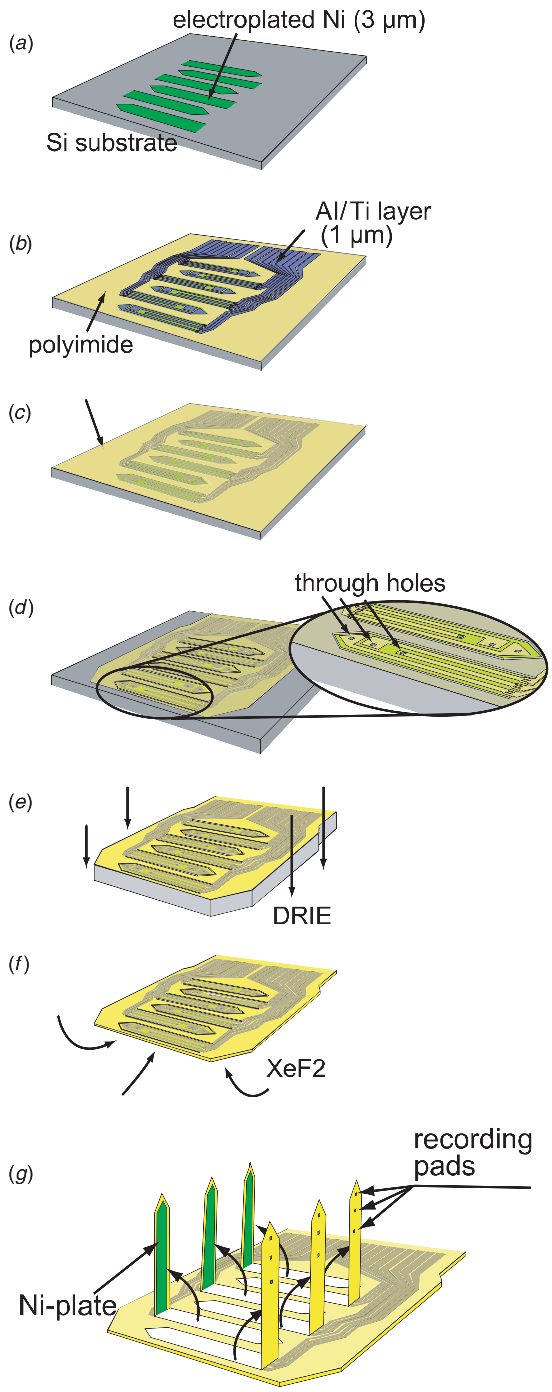

Figure 3. Fabrication process of the flexible probe array. (*a*) Electroplated nickel on a silicon wafer. (*b*) Spin-coat polyimide. Metal layers are deposited and patterned. (*c*) Spin-coat polyimide again to cover the metal layer. (*d*) Pattern the polyimide by O_2 plasma. (*e*) Etch the silicon from top and backside by DRIE. (*f*) Remove the residual silicon on the backside of the electrode by XeF_2. (*g*) Fold the probes.

nickel as the magnetic material. The fabrication details on the structure of the probe are described in the following text (see figure 3).

A 250 μm thick silicon substrate was used as the starting material. Nickel electroplating was done first to form the magnetic backing plate of each probe. The thickness of the nickel plate was 5 μm in this work. A titanium layer was deposited on an aluminum layer with a vacuum evaporator (Sanyu, SVC-700TM) to be used as the electrode pads and interconnections and patterned by wet etching. The total thickness of these metals was about 1 μm. These layers were sandwiched between two 10 μm polyimide layers (HDmicrosystems, PIX3400), which were spin coated and patterned by O_2 plasma for about 30 min. During this

patterning, an aluminum layer was deposited and patterned on the polyimide layer to be used as a mask to define the electrode outline, bonding regions and recording pads. After this process, we formed the through holes, and the titanium layer appeared (figure 3(*d*)). These through holes were formed on each end of the interconnecting cable, so that only this area of the cable would contact the neural tissues. The aluminum mask layer was removed with an etchant (H_3PO_4:HNO_3:CH_3COOH:H_2O = 10:1:1:2) that did not attack the titanium layer.

We used deep reactive ion etching (DRIE) and XeF_2 etching processes to remove the silicon substrate. We first used DRIE to etch about 20 μm from the top, and then again from the backside. Finally, the residual silicon was isotropically etched by XeF_2 to produce the flexible electrode whose total thickness was about 25 μm.

Figure 4 has photos of a 2 × 3 probe array with 18 neural recording channels. Each probe is 1.2 mm in height and 160 μm in width and has three 20 × 20 μm^2 recording pads at 200 μm intervals (figure 4(*d*)). The precise depths of each recording pad when implanted were 0.65 mm, 0.85 mm and 1.05 mm. Since the whole structure was fabricated with flexible materials (polyimide and metal thin films), it was robust and easy to handle. Usually, the wiring from the recording pads to the measuring equipment creates a serious implementation problem. In our process, however, the cables could be integrated with the same surface micromachining technique.

One of the main concerns with this type of flexible probe is that the insertion process is not done effectively. For simplicity, we assumed that the probe was a composite beam of a 20 μm polyimide, and an electroplating Ni plate 5 μm in thickness, 1.2 mm in length and 160 μm in width. Their respective Young's moduli were 2 GPa and 176 GPa [9]. Here, the effective Young's modulus is calculated as 36.8 GPa. Under these conditions, Euler's buckling load can be roughly calculated as about 50 gf, while it decreases to about 1.5 gf without a nickel-backing plate. As shown in figure 4(*c*), each probe can be bent and maintain an upright vertical position due to the nickel-backing plate, which may also be inserted smoothly into organs.

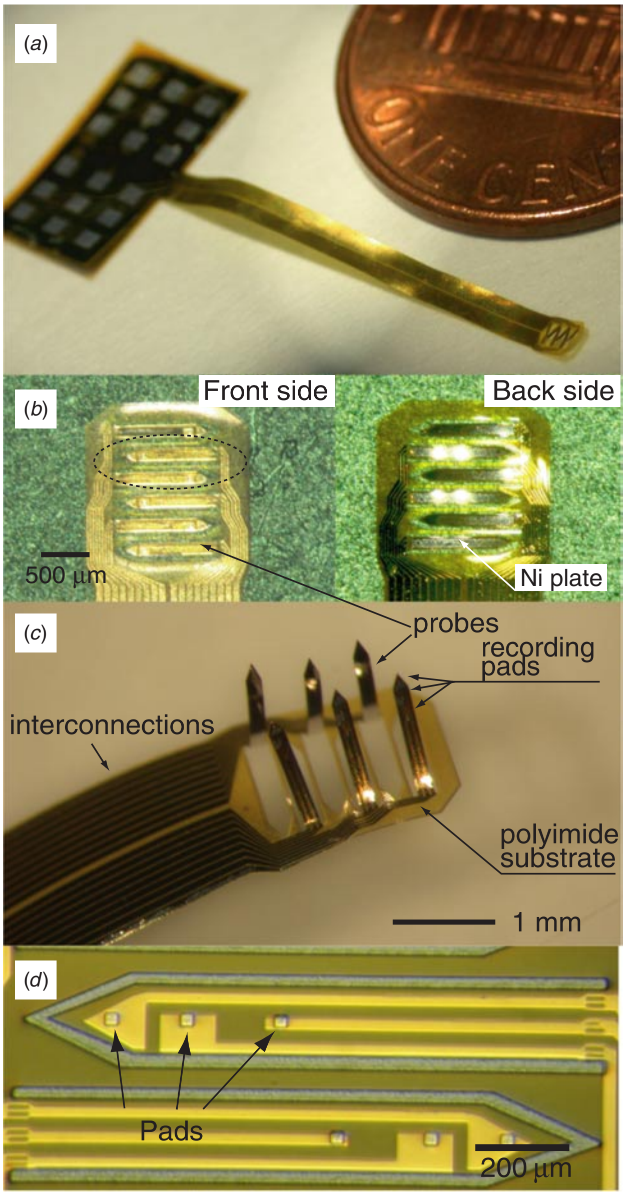

Figure 4. Photos of the flexible multichannel probe array. (*a*) The probe array with 2 cm interconnection cable. (*b*) Front and back side of the probe array before folding. A magnified photo of the dotted area is shown in (*d*). (*c*) 3D flexible probe array after folding. The recording pads are vertically aligned. (*d*) Each probe has three recording pads. They will be bended up after the folding process.

3. Neural recording experiments

The electrode impedance was measured with an LCR meter (Hewlett Packard, 4263B). The results for one recording pad are plotted in figure 5, which shows that impedance is of the order of 1 MΩ at 1 kHz. When all the probes were folded up, the impedance was still unchanged. The hinging parts (i.e., bending parts) of the probe did not break, and electrical conductivity was maintained.

We used an anesthetized rat (Wister rat) for the neural recording experiment for this probe array. It was inserted into the rat's cortex after peeling off the dura mater, which is a tough fibrous membrane covering the cortex. Figure 6 has photos of this insertion experiment. The probes were easily inserted into the organs at an angle, when the polyimide substrate was pushed with tweezers under a microscope. We did not observe probes bending back during the insertions although we did have such a problem when we used the

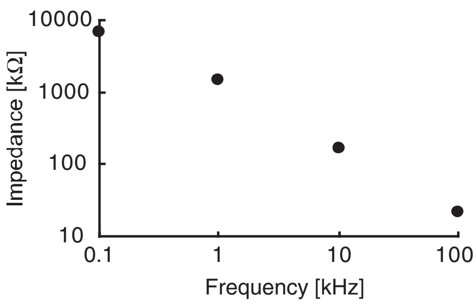

Figure 5. Measurement result of the electrode impedance shows 1.5 MΩ at 1 kHz which is the usual frequency of the neural signals.

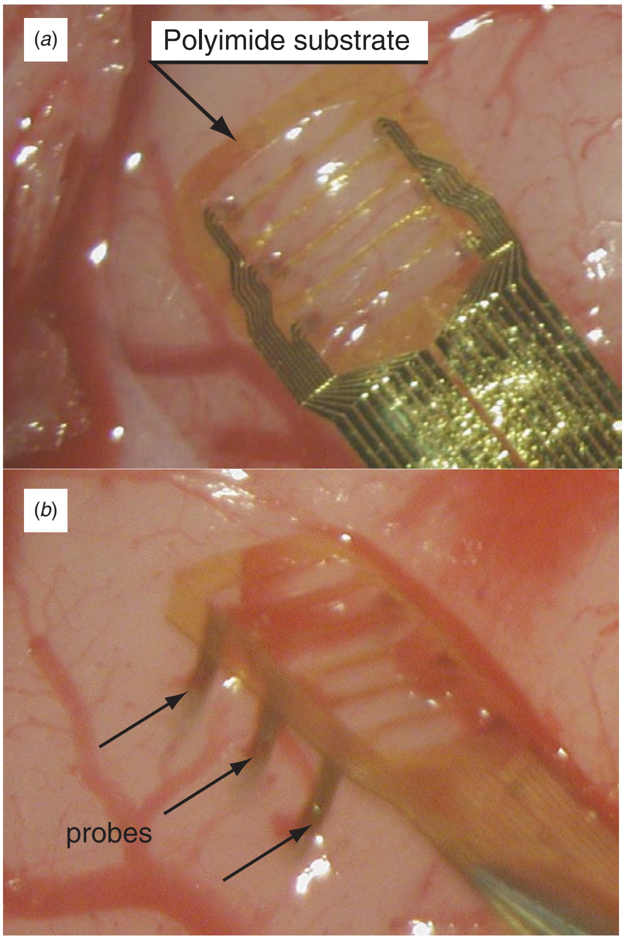

Figure 6. Photos of the insertion experiment. (*a*) The flexible substrate was fit to the brain surface after the electrode insertion to rat's brain. (*b*) The inserted regions can be observed after the detachment of the electrode by tweezers.

20 μm polyimide probe without the nickel-backing plate. This indicates that the nickel-backing plate enhanced the strength against buckling of the probe to improve insertion. However, nickel is known to be toxic and has poor biocompatibility. This kind of problem can be overcome by coating the nickel-backing structure with biocompatible materials. For example, parylene has good biocompatibility [10] and is easy to combine with our fabrication process as we discussed in our previous report [11].

After doing this, the flexible substrate fitted into the brain surface well (figure 6(*a*)). At the end of the experiment, no breakage of the electrode was observed when the electrode was retrieved (figure 6(*b*)). This demonstrated that insertion was repeatable.

Using this electrode, we successfully measured the neural activity from the rat's brain. The neural signals were amplified by a preamplifier (Nihon Koden, JP-640J) and a main amplifier (Nihon Koden, MEG-6116, AB-610J, HPF: 500 Hz, LPF: 3 kHz). They were then recorded on a PC by way of an AD converter at a sampling rate of 10 kHz. Figure 7 has the results of the recorded data from the secondary visual cortex. These indicate that the electrode is useful in 3D multichannel neural recording.

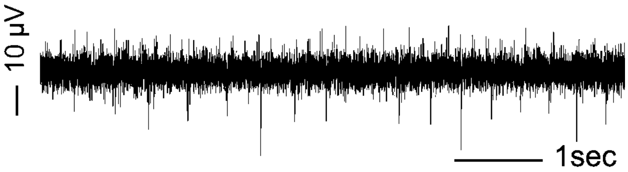

Figure 7. Neural signal was obtained by means of the flexible probe array which was inserted to the rat's visual cortex. Spontaneous action potentials can be observed from the signals.

4. Conclusion

A totally flexible 2 $\times$ 3 probe array was designed, fabricated and tested. The probe was formed with two metal layers sandwiched between a 10 μm polyimide layer. Using a specific process to fold the planar probe structures, we fabricated a 3D probe array with interconnections. Each probe was 1.2 mm long and 160 μm wide, and had three 20 $\times$ 20 μm^2 recording pads at 200 μm intervals. The probe array was inserted into a rat brain and well accommodated by the organs. Neural activity from the brain was successfully measured on a continuous basis.

References

[1] Campbell P K, Jones K E, Huber R J, Horch K W and Norman R A 1991 A silicon-based, three-dimensional neural interface: manufacturing processes for an intracortical electrode array *IEEE Trans. Biomed. Eng.* **38** 758–68

[2] Kovacs G T A, Storment C W, Halks-Miller M, Belczynski C R Jr, Santina C C D, Lewis E R and Maluf N I 1994 Silicon-substrate microelectrode arrays for parallel recording of neural activity in peripheral and cranial nerves *IEEE Trans. Biomed. Eng.* **41** 567–77

[3] Ensell G, Banks D J, Ewins D J, Balachandran W and Richards P R 1996 Silicon-based microelectrodes for neurophysiology fabricated using a gold metallization/nitride passivation system *J. Microelectromech. Syst.* **5** 117–21

[4] Stieglitz T, Beutel H and Meyer J-U 1997 A flexible, light-weight multichannel sieve electrode with integrated cables for interfacing regenerating peripheral nerves *Sensors Actuators* A **60** 240–3

[5] O'Brien D P, Nichols T R and Allen M G 2002 Flexible microelectrode arrays with integrated insertion devices *MEMS'01* pp 216–9

[6] Takeuchi S and Shimoyama I 2000 A three-dimensional shape memory alloy microelectrode with clipping structure for insect neural recording *J. Microelectromech. Syst.* **9** 24–31

[7] Hoogerwerf A C and Wise K D 1994 A three-dimensional microelectrode array for chronic neural recording *IEEE Trans. Biomed. Eng.* **41** 1136–46

[8] Iwase E, Takeuchi S and Shimoyama I 2002 Sequential batch assembly of 3-D microstructures with elastic hinges by a magnetic field *MEMS'02* pp 188–91

[9] Sharpe W N Jr, LaVan D A and Edwards R L 1997 Mechanical properties of LIGA-deposited nickel for MEMS transducers *Proc. Transducer's 97* pp 607–10

[10] Schmidt E M, McIntosh J S and Bak M J 1998 Long-term implants of parylene-C coated microelectrodes *Med. Biol. Eng. Comput.* 96–101

[11] Suzuki T, Mabuchi K and Takeuchi S 2003 A 3D flexible parylene probe array for multichannel neural recording *IEEE Neural Eng.* 154–6