

Silicon Micro-Needles with Flexible Interconnections

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Abstract - A flexible polyimide-based interconnect scheme was developed to realize isolated needle-like microelectrodes. A simple fabrication approach allows the integration of micro-machined silicon needles with a larger silicon base designed to carry elements such as amplifiers, battery or memory. The interconnecting scheme uses two polyimide layers to sandwich a metallic layer. The metal layer forms the electrical connection between the silicon base and the micro-electrodes, while the polyimide layers provide flexible insulation. The current design allows convenient handling of the device during implantation and minimal mechanical load on the implanted region. The device can conform to the surface of neural tissue and allows convenient interfacing with rugged and dynamic tissues. Prototype devices were tested for usability and animal-compatibility. The devices were implanted in sea slugs (*Tritonia diomedea*) and extracellular signals were acquired. *Tritonia diomedea* show full recovery from surgery and implantation, and survive up to a minimum of fourteen days with the ability to perform normal behaviors.

Keywords - Intracellular, flexible, micro-electrodes, needles.

I. INTRODUCTION

Many extracellular MEMS-based microelectrode array designs have been suggested over the past few years [1-9]. These designs offer, among other features, multi electrode recording, compact geometry and on-chip electronics. Many of these designs have used silicon micro-machining. The mechanical strength of silicon was exploited to construct rigid blades with arrays of co-planar metallic electrodes [1,7] as well as silicon needles suited for extra-cellular recording [2]. Recently we have demonstrated the possibility to make silicon needles suitable for intracellular recording using standard silicon micro-machining [10,11].

While rigid devices have established themselves as successful devices for several applications, flexibility is a very desirable property for in-vivo application [3,8]. The focus of the work presented here is to build self-contained implantable devices suited to record intracellular neuronal signals from freely behaving animals. Our technique allows the integration of rigid silicon needles with polyimide (PI) interconnects, creating a flexible link between rigid silicon structures. The semi-independent movement of components linked by flexible interconnects provides the opportunity to partition implant tasks to different locations while accommodating the dynamic nature of soft tissues in animals.

We envisage a device, implanted inside the sea slug with an array of small needle electrodes on the brain, each connected by a polyimide wire to a larger silicon chip where signals recorded by the electrodes are processed and stored (Fig. 1). We use *Tritonia diomedea* as a test bed for this technology for several reasons. Foremost, the brain of this animal has been extensively studied and many individual neuron functions and locations are well known, thus making assessments of implant function possible. Furthermore, the large size (up to 400 μm) and resilience of its brain cells should facilitate our efforts to design, test, and use our new electrodes. In addition, the sea slugs have a large visceral cavity in which we can bury the larger silicon chip/hardware. Ultimately, we wish to use these arrays of flexibly connected electrodes to process neuronal signals from multiple cells and brain locations, while the animal's behavior is monitored. This will lead to a much deeper understanding of how the brain controls behavior through the activity of both individual and groups of neurons. The technique presented here is a necessary step toward devices suited for intracellular recording from freely behaving animals.

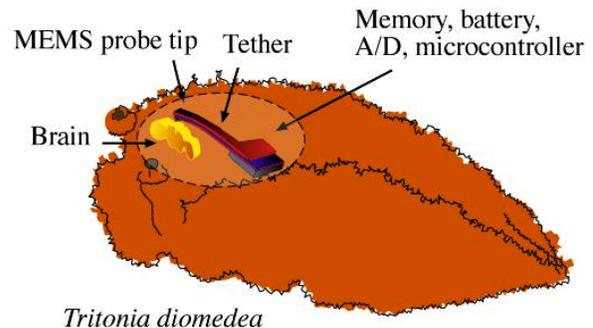


Fig. 1. A schematic drawing of a *Tritonia diomedea* with implanted device. Typical animal length is 10 cm. Neurons can be as large as 400 μm . Overall brain dimensions are several mm.

In the following sections we describe the general device designs used in this work. The details of the fabrication process are presented. Usability and preliminary recording results are presented and discussed.

II. FABRICATION CONCEPTS

We have previously shown that standard micro-machining techniques can be employed to realize arrays of sharp

silicon needles suited for intracellular neuronal recording (Fig 2). The needle process consists of initial tall silicon pillars and a sequential reactive ion etching (RIE) SF_6 sharpening [10]. To manipulate these electrodes, each needle had to be individually handled by its silicon base. In doing so, we encountered several difficulties: Handling each needle electrode is difficult and time consuming and establishing electrical contact with the needle was cumbersome.

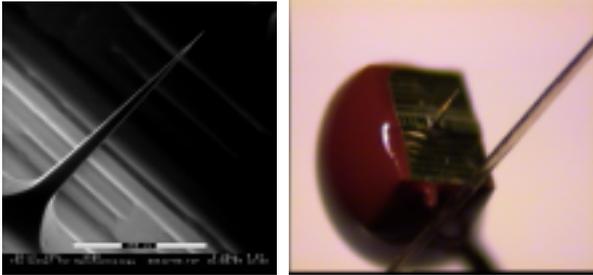


Fig. 2. (Left) ESEM image of a silicon needle. Scale bar is 150 μm . (Right) Manually wired needle shown relative to human hair. The needle was wired to a thin insulating metallic wire. The contact was passivated with varnish.

The current design procedure was motivated by our desire to achieve recording from inside cells using silicon needles similar to those shown in Fig. 2 [11]. To overcome the handling difficulties mentioned above, the needles in our new method are pre-wired and in addition, a full implant design is incorporated into a single integrated process.

The main components of this device (Fig. 3) consist of a large silicon base, small silicon needles, and a connecting metallic wire encapsulated between two polyimide layers (Polyimide 1 and 2). Polyimides have been extensively investigated and demonstrated useful for similar applications [8,12]. The first polyimide film (polyimide 1) has a via, directly above the base of the silicon needle. The metal film (aluminum) is deposited directly on top of the first polyimide film and thus contacts the silicon of the needle base through the polyimide window. The second polyimide layer (Polyimide 2) encapsulates the metal film and overlaps the perimeter of the first polyimide film, except for a reduced base dimension providing a region for subsequent electrical connection to the aluminum layer. The result is a flexible, insulated electrical connection between the silicon needle and the large aluminum pad supported by a silicon base.

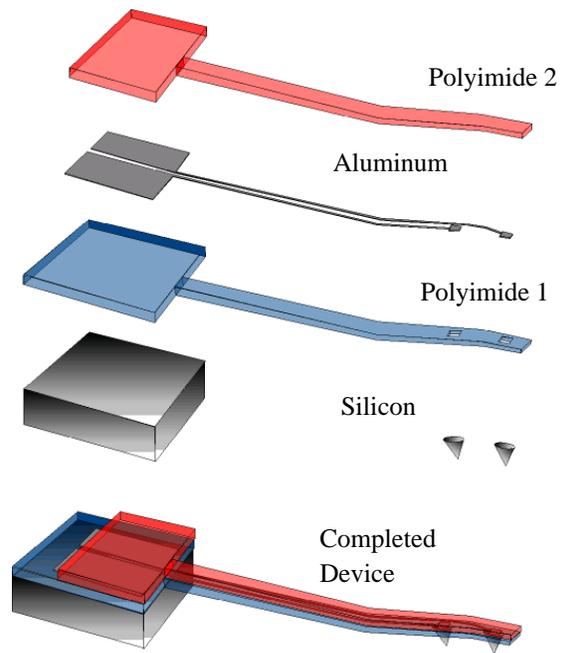
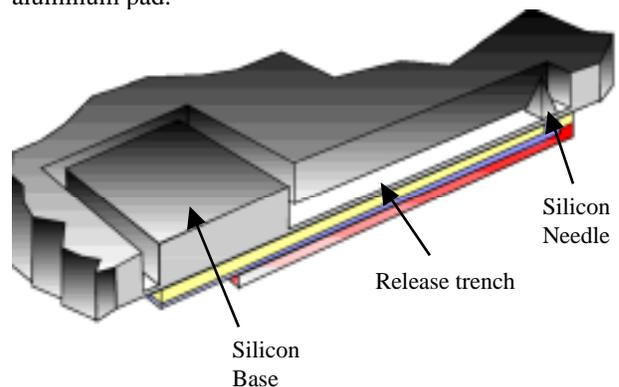


Fig. 3. Building blocks of final device showing encapsulating (polyimide-aluminum-polyimide) film layers attached to silicon needles and the larger silicon base.

The realization of this structure is achieved by a process schematically presented in Fig. 4: Deep trenches are etched in a silicon wafer. These trenches define the shape of the silicon parts in the final device (silicon base and silicon needle) and are used to release the device from the remaining wafer. Fig. 4 represents one intracellular device prior to final release with a single sharpened silicon pillar. The figure (top) shows the backside release trenches, which define the final device shape. The thin silicon (approximately 20 μm) channel floor allows for a frontside planar surface to process PI films (bottom, Fig. 4). During the final etch-release step, the silicon channel floor is completely etched. Simultaneously the small silicon pillar is sharpened into a needle. As a result the base and the needle are connected via the metal layer with the flexible PI films. With adequate passivation on the needle electrode base such a device would allow intracellular recording through the sharpened needle with a connection made to the exposed aluminum pad.



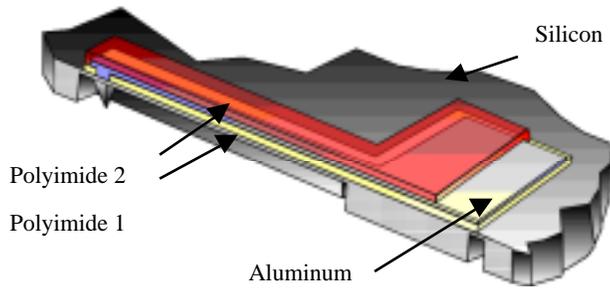


Figure 4: A schematic drawing of flexible neural implant device developed to allow recording from freely behaving animals. The devices consist of a large silicon base, a metallic wire encapsulated between two polyimide (PI) layers and small silicon needle. (Top) Backside view. (Bottom) Front-side view.

The process is not limited to any specific geometry. Many complex patterns of polyimide wiring can be easily designed (Fig. 5). We are thus free to create shapes which provide the desired dimensions, flexibility, electrode array geometry, and device density on the source silicon wafer. For example, in our sea slug experiments thus far we have used the elbowed pattern (Fig. 5, right). The 45° bend reduces the stress on the silicon base when manipulating the electrodes during implantation. In addition, after the smaller needle chip is glued to the brain, the larger silicon base is positioned in an area convenient for implantation.

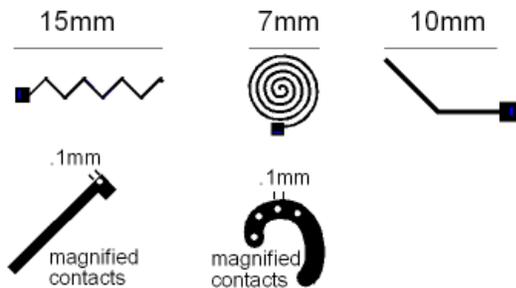


Fig. 5. Examples of possible electrode designs.

III. FABRICATION

The fabrication process involves the following steps (Fig. 6): First thermally oxidized conducting (4", 400 μ m, p-type, double sided polished) silicon wafer was patterned with thick AZ4620 (Clariant) photoresist (10 μ m) to define release trenches on the wafer backside (Fig. 6a). The wafer front side was protected using blue semiconductor tape and the backside unmasked SiO₂ regions were etched in buffered oxide etch (BOE). The frontside was protected from SiO₂ etching to insure a polished frontside after deep reactive ion etching (DRIE). Next, the wafer was introduced into a DRIE chamber (Oxford Instruments) and etched using the Bosch etching process (Fig. 6b). After DRIE etch, the wafers were cleaned in cleaning solvent (EKC830, EKC Technology, Inc.) to remove DRIE residual polymer and mask material,

rinsed and dried. Next, a polyimide layer was patterned and cured (10 μ m after cure) on the opposite side of the wafer (Fig. 6c). The alignment to the backside release trenches was accomplished using an IR ABM aligner. It was critical to fully cure the first polyimide film prior to further processing as the polyimide strength and adhesion would be compromised. Also, since multiple films were to be processed, a lower final temperature cure was used (350C). Following the first polyimide layer, a metal was deposited (5 μ m aluminum, E-beam) for ohmic contact to the silicon needle base. To insure continuity, a blanket metal deposition was completed and photoresist patterned with a clear field mask. The unmasked aluminum regions were etched (Alameda Chem). Prior to the second polyimide film application, the base polyimide and metal layers were introduced to oxygen plasma to insure maximum adhesion. The second polyimide-encapsulating layer was processed similarly albeit at higher final temperature (450 °C).

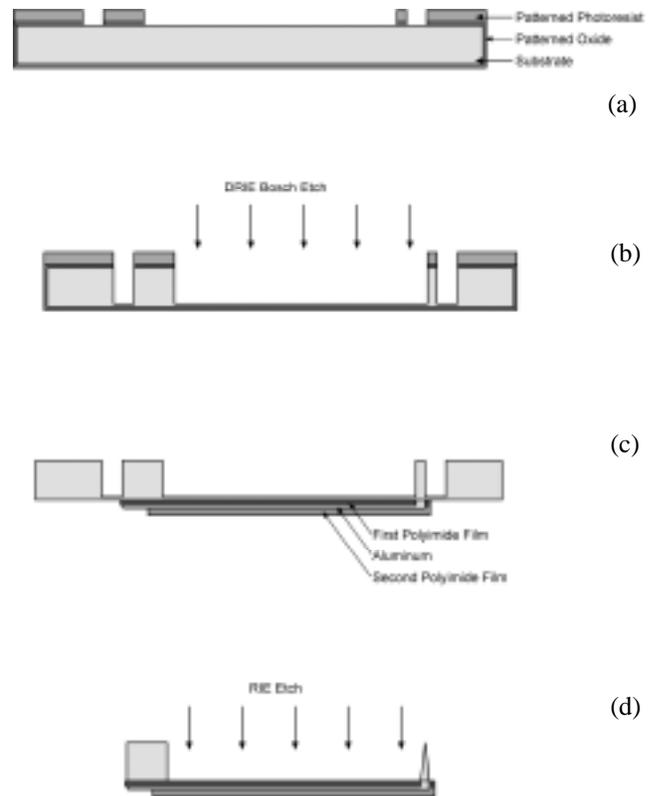


Fig. 6. Process sequence for the flexible electrode design. (a) AZ4620 photoresist mask (b) DRIE etching of the trench, a thin silicon layer supports the remaining structure through the entire fabrication process (c) Two polyimide layers and an embedded metal layer are deposited and patterned on the surface. (d) A final release process removes the supporting thin silicon layer and sharpens the silicon pillar onto a needle.

To release the entire structure the wafers were etched (SF₆) in a standard RIE chamber. Finally, the structures were nudged out of the wafer (Fig. 6d). Therefore, the DRIE etch mask design was designed to compensate for the slower etch rate surrounding the base and pillars. The final SF₆ etch process was also used to sharpen the silicon pillars into

needles suitable for intracellular recording. The parameters of our current process are designed to achieve the release and the sharpening using the same RIE step.

An identical process flow can be used to achieve flat metallic electrodes with flexible interconnects to a rigid silicon base. A slight modification in the design, namely omitting the pillar in the etch mask, exposes the metal through the base polyimide film during the release process. Structures without silicon needles are useful for extracellular recording experiments. Such devices with gold electrodes were fabricated.

IV. RESULTS

Twenty-five electrodes were batch fabricated and released on each 4" wafer (Fig. 7). Both sharpened silicon needles and bare metal patch electrodes were successfully manufactured. We confirmed that both types of devices were electrically connected to the base metal contact area. Tests involving stepwise immersion in electrolytes demonstrated that the polyimide encapsulation effectively insulated the metal areas between the base contact and the electrode.

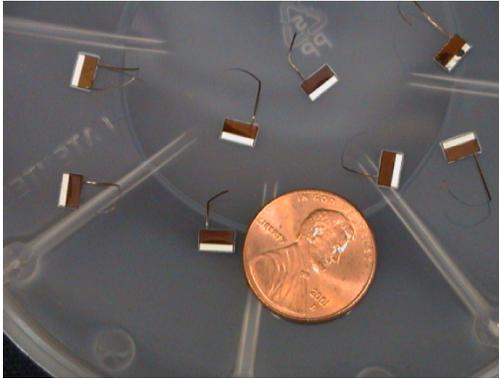


Fig. 7. Released, flexible silicon devices ready for implantation. (Top). The same batch includes both sharpened silicon pillars at the end, (one enlarged below) and devices with bare metal patch electrode.

Extracellular patch electrode devices were used for implantation studies since the recording abilities of the

silicon needle electrodes need further refinement. The implant was first prepared for implantation by attaching an ultra fine wire (Copper/Beryllium alloy, 25 μ m diameter California Fine Wire, Inc) to the base metal contact. This contact was then insulated using silicone surgical glue (Kwik Sil, World Precision Instruments, Inc). The sea slug was anaesthetized, and an incision near the brain was cut. This incision was held open by hooks to expose the sea slug brain and visceral cavity (Fig. 8). The patch electrode was surgically glued (Nexaband S/C, Veterinary Products Laboratories) overtop of a neuron. After curing, the insulated silicon base was tucked inside the visceral cavity. An additional fine wire (reference electrode) was glued in close proximity (1-2 mm) to the test electrode device.

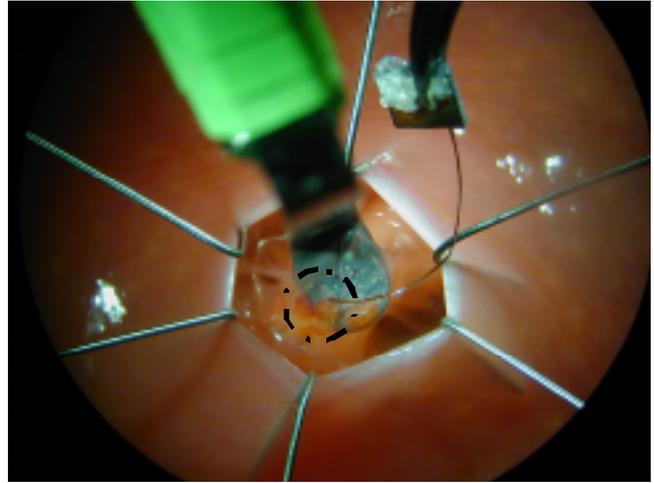


Fig. 8. Implantation procedure: the metal patch electrode is placed on the exposed brain of a sea slug and the silicon base with the external wires was tucked under the slug's skin.

After successful implantation, the incision was glued shut (Nexaband Fluid, Veterinary Products Laboratories) and the fine wire leads were attached to a differential amplifier. The animal was released into a tank, free to crawl around with the fine wire trailing along the water surface above it.



Fig. 9. A sea slug with implanted device several hours after surgery. The slug can freely move in the water tank. The slugs can survive more than two weeks after such a surgery procedure.

Extracellular signals were recorded over the next four days. Typical results are shown in Fig. 10. The differential signal shows clear biological signals.

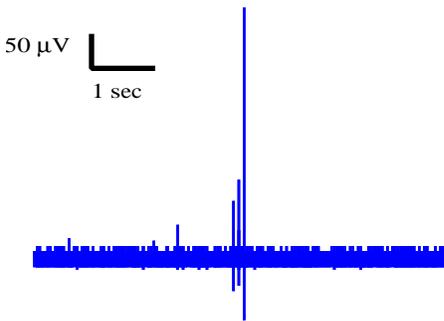


Figure 10: Six second snapshot of extracellular cell signals using a metallic patch electrode at the end of a flexible polyimide wire.

V. DISCUSSION

The fabrication process described in this paper is quite straightforward. However, several issues should be noted. The final etch step is accountable for both the needle sharpening and the final removal of the silicon trench. If the process is not tuned properly the needles may not be sharpened by the time the trench is removed. Since degradation of the constructed PI films during the final etch step is unavoidable, prolonged RIE etching may damage the PI or overheat it. At the other extreme, the pillar may be over-etched by the time the trench is removed. To minimize these effects the DRIE etch rate has to be precisely monitored. Geometrical factors on the etch rate have to be accounted for in the mask design.

Similar results to those described here can be obtained by reversing the fabrication process by first performing PI and metal patterning and only then performing the DRIE process. Even though we have managed to perform such a process, it should be noted that this process is inferior due to DRIE polymer residue interference with the final needle sharpening. To sufficiently remove the film, aggressive solvents (EKC830) are necessary and may affect polyimide adhesion and/or permeability (affecting insulation properties). The obvious solution to this problem was to complete the backside bulk etch first. The only concern was processing the remaining films on the fragile substrate (made more fragile by the backside etched release trenches). Care was taken to not etch through the wafer completely. An additional factor of importance related to the DRIE etching concerns the etch profile. To successfully sharpen the needles it was necessary that the pillar sidewalls were nearly vertical. This was achieved by fine tuning the etching parameters and the mask design.

The durability of polyimide based devices in harsh environment is limited [3] and may not be adequate for long implantation time required for human implants. For the application discussed here, two day durability is sufficient and is completely compatible with expected PI coating performances. In our preliminary tests, devices after several days of use were still intact.

Future work will focus on improving needle characteristics, including: Sharpening, conductivity and passivation. Future experimentation will target specific neurons in comparison with standard recording methods, longer term recording, and initial attempts to correlate behavior with neuronal activity in freely behaving animals.

VI. SUMMARY

We have described a new scheme to interconnect silicon needles using micro-fabricated flexible polyimide based wires. Our design creates an electrical connection between a larger silicon chip and smaller electrodes at the end of a polyimide wire. The result is an easily handled implant, with a strong, yet flexible connection to either a silicon needle or metallic patch electrode, which can be used for neuronal recording experiments. Initial usability tests demonstrated the effectiveness of the method to manufacture devices which are easy to use and handle. The versatility of the design is particularly useful for attempts to comply with varied user requirements. Preliminary extracellular recording with an exposed gold electrode demonstrate the usability of these devices.

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REFERENCES

- [1] D.J. Anderson, K. Najafi, S.J. Tanghe, D.A. Evans, K.L. Levy, J.F. Hetke, X. Xue, J.J. Zappia, K.D. Wise, Batch fabricated thin-film electrodes for stimulation of the central auditory system, *IEEE-Transactions-on-Biomedical-Engineering*, 36, 693–704 (1989).
- [2] P.K. Campbell, K.E. Jones, R.J. Huber, K.W. Horch, and R.A. Normann, A silicon-based, three-dimensional neural interface:

- manufacturing processes for an intracortical electrode array, *IEEE-Transactions-on-Biomedical-Engineering*, 38, 758-768 (1991).
- [3] J.F. Hetke, J.L. Lund, K Najafi, K.D. Wise, and D.J Anderson, Silicon ribbon cables for chronically implantable microelectrode arrays, *IEEE-Transactions-on-Biomedical-Engineering*, 41, 314-321 (1994).
- [4] P.J. Rousche, D.S. Pellinen, D.P. Jr. Pivin, J.C. Williams, R.J. Vetter, D.R. Kirke, Flexible polyimide-based intracortical electrode arrays with bioactive capability, *IEEE-Transactions-on-Biomedical-Engineering*, 48, 361 -371 (2001).
- [5] N.A. Blum, B.G. Carkhuff, H.K. Jr. Charles, R.L. Edwards, R.A. Meyer, Multisite microprobes for neural recordings, *IEEE-Transactions-on-Biomedical-Engineering*, 38, 68 -74 (1991).
- [6] J.J. Mastrototaro, H.Z. Massoud, T.C. Pilkington, R.E. Ideker, Rigid and flexible thin-film multielectrode arrays for transmural cardiac recording, *IEEE-Transactions-on-Biomedical-Engineering*, 39, 271 -279 (1992).
- [7] G.T.A. Kovacs, C.W. Storment, M. Halks-Miller, C.R. Jr. Belczynski, C.C.D. Santana, E.R. Lewis, N.I. Maluf, Silicon-substrate microelectrode arrays for parallel recording of neural activity in peripheral and cranial nerves, *IEEE-Transactions-on-Biomedical-Engineering*, 41, 567 -577 (1994).
- [8] D.P. O'Brien, T.R. Nichols, M.G. Allen, Flexible microelectrode arrays with integrated insertion devices, *Proceedings of the 14th IEEE International Conference on Micro Electro Mechanical Systems*, 2001, 216 -219.
- [9] S. Metz, F. Oppliger, R. Holzer, B. Buisson, D. Bertrand, P. Renaud, Fabrication and test of implantable thin-film electrodes for stimulation and recording of biological signals, *Proceedings of the 1st Annual International Conference On Microtechnologies in Medicine and Biology*, 2000, 619 -623.
- [10] Y. Hanein, K.F. Böhringer, R.C. Wyeth, and A.O.D. Willows, Towards MEMS Probes for intracellular recording, *Sensors Update* 10, 1-29 (2002).
- [11] Y. Hanein, U. Lang, J. Theobald, R. Wyeth, K.F. Böhringer, T. Daniel, D.D. Denton, and A.O.D. Willows, Intracellular recording with high aspect ratio MEMS neuronal probes, *Proceedings of the International Conference on Solid-State Sensors and Actuators (Transducers)*, 2001, Munich, Germany.
- [12] D. Y. Shih, H. Yeh, C. Narayan, J. Lewis, W. Graham, S. Nunes, J. Paraszczak, R. McGouey, E. Galligan, J. Cataldo, R. Serino, E. Perfecto, C. A. Chang, A. Deutsch, L. Rothman, J. Ritsko, J. Wilczynski, Factors affecting the interconnection resistance and yield in the fabrication of multilayer polyimide/metal thin film structures, *Proceedings of Electronic Components and Technology Conference*, 1992, 1002 -1014.