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Electrostatic Microactuators for Precise Positioning of Neural Microelectrodes

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Abstract

Microelectrode arrays used for monitoring single and multineuronal action potentials often fail to record from the same population of neurons over a period of time likely due to micromotion of neurons away from the microelectrode, gliosis around the recording site and also brain movement due to behavior. We report here novel electrostatic microactuated microelectrodes that will enable precise repositioning of the microelectrodes within the brain tissue. Electrostatic comb-drive microactuators and associated microelectrodes are fabricated using the SUMMiT V™ (Sandia's Ultraplanar Multilevel MEMS Technology) process, a five-layer polysilicon micromachining technology of the Sandia National labs, NM. The microfabricated microactuators enable precise bidirectional positioning of the microelectrodes in the brain with accuracy in the order of 1 μm . The microactuators allow for a linear translation of the microelectrodes of up to 5 mm in either direction making it suitable for positioning microelectrodes in deep structures of a rodent brain. The overall translation was reduced to approximately 2 mm after insulation of the microelectrodes with epoxy for monitoring multiunit activity. The microactuators are capable of driving the microelectrodes in the brain tissue with forces in the order of several micro-Newtons. Single unit recordings were obtained from the somatosensory cortex of adult rats in acute experiments demonstrating the feasibility of this technology. Further optimization of the insulation, packaging and interconnect issues will be necessary before this technology can be validated in long-term experiments.

Keywords

Brain implants; MEMS; microdrive; multi-unit activity; prostheses

I. Introduction

CURRENT microelectrode technologies that record from a population of neurons simultaneously using arrays of multichannel microelectrodes enable us to understand the emergent properties of assemblies of neurons [1]–[6]. These are especially useful when combined with simultaneous measures of behavioral performance as pretrained animals perform sensory-perceptual tasks. It is, therefore, critical during these studies that measurement of electrophysiological parameters do not hinder or distort the behavioral performance or vice

versa. The early, fixed or unmovable electrode arrays were produced by photolithography [7]-[14]. One of the shortcomings of this type of fixed electrode arrays was the inability to position each electrode individually. Therefore, one was constrained to position the entire array of electrodes at an optimum position that will maximize the number of neurons that could be recorded. This lack of flexibility also affected the spatial precision of the measurement. In addition, during chronic measurements in freely moving animals, it is common to experience degradation in the electrical activity from individual recording sites [15]. It is hypothesized that the loss of electrical contact could be due to a gradual drift in the microelectrode or tissue movement within the animal or glial encapsulation of the recording site. The neuron could also be lost due to impalement by the electrode or death due to pathological reasons. These limitations on the fixed electrode arrays have made electrophysiological recordings over long periods of time with implanted electrodes, very difficult and cumbersome.

A slightly different approach was used in the Reitboeck method of multielectrode recording [16]-[19]. This method allowed independent movement of each electrode. The individual electrodes could also be arranged in a choice of spatial patterns in X - Y dimensions. The current version of this electrode and the associated microdrives are produced commercially by Uwe Thomas Recording (Marburg, Germany). It allows a shock-free positioning of quartz-platinum microelectrode in 1 μm steps. The micromanipulator is powered by DC micromotors that allows penetrations of the dura and recordings from structures at depths down to 20 mm. However, this recording system would still not be suitable for recording from awake, freely behaving animals that are allowed to move around. Frederick Haer and Co. (Bowdoinham, ME) and Alpha Omega Engineering (Nazareth, Israel) have also developed similar microelectrode positioning systems all of which can only be used when the animal's head is restrained.

Wilson and McNaughton [20] have successfully recorded from the pyramidal layer of CA1 in the hippocampus of awake, freely moving rats. Using implanted microelectrode arrays containing tetrodes, they were able to record from as many as 148 hippocampal neurons in some cases. Each microelectrode could be independently adjusted *manually* with a precision of approximately 10 μm . The array weighed about 10 g and did not significantly restrict the free movement of the animal. A modern version of a chronically implantable microdrive system (14 channels), developed by David Kopf Instruments (Tujunga, California) provides independent adjustments of 5–10 μm in each channel with a total electrode movement of 7.5 mm. However, the *manual* adjustment of the electrode arrays in the above methods makes them imprecise and hard to replicate. A similar manual microdrive technology for chronic neuronal recordings in behaving monkeys has been reported recently in [21]. Microdrive technologies using commercially available micro-motors have also been successfully developed and tested in semi-chronic experiments [1], [15] further confirming the effectiveness of actuation in improving the single neuronal recording capabilities in chronic experiments.

We report here a novel micro-electromechanical systems (MEMS)-based electrostatic microactuator technology that will enable precise independent positioning of the individual microelectrodes in acute experiments. We report the successful development and initial test results of the microactuated microelectrode technology in acute rodent experiments. The small size of the microactuated microelectrodes and the integrated, batch-processing capabilities in fabrication will be added advantages compared to the existing microdrive technologies. However, the effectiveness of this technology in chronic experiments will have to be tested in future studies. The technology reported here can be readily extended to an array of microelectrodes on a single substrate. In addition, the process of microfabrication lends itself readily to future integration with signal conditioning circuitry on the same substrate.

II. Materials and Methods

A. Principle of Electrostatic Comb-Drive Microactuator

In an electrostatic comb-drive microactuator, the force generated among interdigitated comb structure fingers is used to drive gear trains and transmissions. The fabrication methods and testing results for this actuation system and the systems enabled by this microengine have been reported by Sandia National Laboratories (Albuquerque, NM) [22]-[24]. The electrostatic attraction due to the fringing field between the moving and stationary fingers in this comb-structure provides the linear actuation when voltage is applied [Fig. 1(a)]. Springs fabricated in the same polysilicon layers as the fingers of the comb-structure provide the restoring force when the voltage is removed. A schematic of a subassembly of the electrostatic actuator showing a bank of parallel plates is shown in Fig. 1(b). The linear motion of two such actuators (x - and y -drives) is transformed into a rotational motion as illustrated in Fig. 1(c) and (d).

B. Design

Design and fabrication of the devices was done using several layers of polysilicon in the standard SUMMiT V™ (Sandia's Ultra-planar Multilevel MEMS Technology) technology. The schematic of a typical five-layer polysilicon (*poly0* through *poly4*) SUMMiT V™ process is illustrated in Fig. 2. The overall design of the microactuator and the associated microelectrode is shown in Fig. 3(a) and (b), respectively. The detailed rendering of the microelectrode tip in Fig. 3(b) shows the different polysilicon layers in the microelectrode and the guide rails above them. A schematic of how the microactuated microelectrode was eventually used in the rodent is shown in Fig. 3(c) using a cartoon of the coronal section of a rodent brain. There were two different designs, one where the comb drive was directly attached to the neural microelectrode [Fig. 4(a)], and the other with two torque amplification stages of 12 : 1 gear ratio between the comb drive and the neural microelectrode [Fig. 4(b)]. These were components in the standard parts library of SUMMiT V™ technology. The comb drive teeth were placed in *poly3* layer while *poly1* and *poly2* layers were laminated to form the lower layers of the microelectrode. Guides were placed above the microelectrode and a pin-in-groove alignment structure was placed toward the front of the microelectrode as shown in Fig. 5. *Poly0* layer was used as a ground plane underneath all the structures.

C. Fabrication

A brief description of the SUMMiT V™ process is given here, and the reader is referred to [23], [24] for more information. Silicon wafers (6 in) were used as starting substrate in all cases. Insulation layers, 0.6 μm oxide and 0.8 μm low stress nitride were grown and deposited, respectively. Cuts performed in this layer provided electrical contacts to the substrate. *Poly0* layer (0.3 μm polysilicon, 40 Ω/square) was deposited and patterned. A 2- μm -thick sacrificial oxide layer (*sacox1*) was deposited; in two etch steps, dimples (partially etched through the oxide) and through holes were defined. Dimples were used to prevent stiction and reduce contact area between mechanical layers. Next, *poly1* layer (1 μm polysilicon, 22 Ω/square) was deposited. Pin-joints and hubs were defined at this stage by first etching the polysilicon and performing an isotropic oxide etch to partially remove the oxide underneath the parts. A conformal (sacrificial) 0.5 μm oxide (*sacox2*) deposition defines the separation between the pins, gears and the hubs. Structures in *poly1* were defined by the next etch step. *Poly2* deposition (1.5 μm , 18 Ω/square) forms the captive mechanical elements (pins, hubs, etc.) when polysilicon conformally fills in the open areas (Fig. 2). Structures (gears, microelectrode) and etch release holes were defined in the etching step. Sacrificial oxide layer, *Sacox3* was deposited and chemi-mechanically polished (CMP) to 2 μm thickness over *poly2* layer. Dimples were defined by etching through the oxide and backfilling with a 0.3 μm thick oxide deposition. Through holes were defined by the next oxide etch step. *Poly3* layer (2.25 μm , 9 Ω/square) was deposited and patterned. Sacrificial oxide layer, *Sacox4* and *poly4* layers

followed the same steps as *sacox3* and *poly3*. Parts were then separated, released in an HF solution, super-critical CO₂ dried and vapor phase SAM (self assembling monolayer) coated. Released parts were attached and wire-bonded in modified DIPs or other packages that had one edge open to allow extension of the neural microelectrode. A micrograph of the entire assembly of microactuator and a microelectrode is shown in Fig. 6.

D. Electrical Contact

The polysilicon microelectrode was moved approximately 3 mm off the chip. A fine polyimide coated stainless-steel wire (California Fine Wires, Grover Beach, CA), 12 microns in diameter was cut to length so as to be sufficiently big to make contact with both the PC board and the microelectrode. Both ends of the wire were stripped of the insulation by burning them under a flame, just enough to expose the conductive core. Silver conductive paste (Electrodag 477SS RFU, Acheson Colloids Company, Port Huron, WA) was applied to both ends of the wire. Electrical contact is made with the electrode near the region closer to the edge of the chip. The other end of the wire was pasted to a bond pad on the PC board that led to an external connector. The chip was then baked in the oven at 120 °C for 15 min.

A simple continuity test was done. The chip was mounted on a manual micromanipulator with the tip of the electrode facing down. A beaker full of saline solution was placed under the chip, without the surface of the solution making any contact with the electrode tip. A simple circuit was connected with the microelectrode, saline solution, an audio frequency generator and a head-phone in the loop. The only break in the circuit was the absence of any contact of the electrode tip with the surface of the solution. The chip was lowered down toward the surface using the micromanipulator. As the electrode tip touched the surface of the solution the change in audio frequency on the head-phone was used as an indicator of electrical continuity.

E. Insulation

The microactuator chip was mounted on a hydraulic micro-manipulator with the electrode tip facing down. A small beaker filled with biocompatible epoxy was placed under the chip. The electrode was steadily dipped into the epoxy till the edge of the PC board, ensuring no epoxy climbed the surface of the chip itself. The chip was then pulled out at the same rate. To partially cure the epoxy the chip was baked in the oven at 99 °C for 30 min. The chip was again mounted on the hydraulic micro-manipulator with the electrode tip facing down. A beaker filled with dichloro-methane was placed below the electrode tip. Dichloro-methane etched the epoxy present on the tip, exposing the conductive core. The electrode was steadily lowered till the tip was just touching the surface (this was done visually by adjusting a Leica GZ6 stereo-microscope horizontally). The electrode was then dipped in exactly 50, 100, or 200 µm depending on the length of exposed tip desired. The electrode was then pulled up. The chip was baked in the oven at 200 °C to fully cure the insulation epoxy and expose a recording site at the tip of the microelectrode. The approximate thickness of the resulting epoxy insulation was 5 µm as estimated under the microscope. The chip housing the microactuator and the microelectrode after insulation and establishing electrical contact is shown in Fig. 7(a). One of these chips is placed adjacent to a US penny for size comparison in Fig. 7(b).

F. Electrical Impedance Test

The chip was mounted on a hydraulic micro-manipulator with the electrode tip facing down. A small beaker filled with phosphate buffered saline (PBS) was placed below the chip. An electro-chemical workstation (Model 660A CH Instruments, Austin, TX) with three terminals was utilized to carry out the 3-electrode impedance test. The first terminal was a platinum counter electrode. The second terminal was a silver wire used as a reference electrode. The third terminal was connected to the electrical pin out of the polysilicon electrode on the chip. The counter and the reference electrodes were positioned inside the beaker filled with PBS,

ensuring no contact was present directly between them. The polysilicon electrode tip was then lowered till the electrode is at least 1 mm inside the solution. An ac impedance test was run from frequencies 0.1 Hz–100 KHz at a voltage of 0.05 V.

G. Surgery and Implantation

Male Sprague Dawley rats of approximately 300 grams were obtained from Arizona State University Animal Care. All procedures were carried out with the approval of the Institute Animal Care and Use Committee (IACUC) of Arizona State University, Tempe. The experiments were performed in accordance with the National Institute of Health (NIH) guide for the care and use of laboratory animals (NIH publications no. 80–23) revised 1996. All efforts were made to minimize animal suffering and to use only the number of animals necessary to produce reliable scientific data, and to utilize alternatives to *in vivo* techniques, if available.

Each animal was initially anesthetized using 1 ml/kg of anesthetic cocktail composed of 100 mg/ml Ketamine, 20 mg/ml Xylazine, 10 mg/ml Acepromazine mixed with sterile water. A maintenance dose of 0.15 ml of the cocktail was administered when necessary. After being anesthetized the animal was then prepared for surgery by shaving the head starting from right in front of the eyes to the base of the neck. After shaving was complete, alcohol and betadine were used to sterilize the surgical area. A skin incision was done to expose the skull. A 1/16th in hand drill was used to bore a hole 1 mm posterior and 4 mm lateral to the bregma. Once the implant site was cleared of bone chips, the dura was cut and partially resected using a 25-gauge needle and the brain was exposed. The electrode was then implanted into the somatosensory cortex (approx. 0.5 mm posterior and 3 mm lateral to the bregma). After implantation, the site was covered with gel foam. Ground electrode was located over the neck muscle.

H. Actuation

The electrostatic comb microactuators were actuated using a 4-channel Super Microdriver™ (Pragmatic systems, San Diego CA) in three phases—acceleration, constant velocity and deceleration. Typical waveforms used simultaneously in all 4 activation sites (*up*, *down*, *left*, and *right*) for actuating the comb microactuator are illustrated in Fig. 8. The *up* and *down* actuation sites are used to move the *x*-drive back and forth and the *left* and *right* actuation sites are used to move the *y*-drive. The waveforms corresponding to *up* and *down* actuation terminals constitute the two halves of a rectified sinusoidal waveform as shown in Fig. 8. The same is true of the waveforms corresponding to the left and right actuation sites. In addition, there is a 90° phase difference between the sinusoidal waveform given to the *up* and *down* actuation sites and that which is given to the *left* and *right* actuation sites. The waveforms in the acceleration and deceleration phases were used respectively to move the microelectrode from rest to a constant rate or bring the microelectrode moving at a constant rate to rest.

I. Data Acquisition and Analysis

Multiunit data was acquired using a 32-channel neural data acquisition system (TDT Inc., Boca Raton FL). Signals were then sorted and analyzed using the software in Plexon multichannel neural acquisition systems (Plexon Inc., Dallas, TX).

III. Results

The rotating gear directly connected to the *x*-drive and *y*-drive actuators is 36 μm in radius that leads to a linear translation of approx. 227 μm for every one cycle of activation of the *x*-drive and *y*-drive actuators. In the case of the design with two torque amplification stages of 12 : 1, the effective linear translation of the shuttle for every one cycle of activation of the *x*-drive and *y*-drive actuators is 1.6 μm . The *x*-drive and *y*-drive actuators can also be powered by any

fraction of a single cycle of the rectified optimized sinusoidal waveform shown in Fig. 8 that leads to a corresponding fraction of the linear displacement due to activation by one complete cycle. However, there is an initial error of approximately 1 μm every time we cycle through one acceleration, constant rate and deceleration phase since the rotating gears have to first engage the moving shuttle. Therefore, while the precision of the microactuators is very high, the accuracy of the linear displacement is limited to approximately 1 μm around the true or desired displacement. The reliability, electrical and mechanical instabilities, and other failure modes of the electrostatic microactuator have been reported earlier [25].

Before insulation of the microelectrode, the electrostatic microactuator can move over a distance of 5 mm. However, since the microelectrode was dip-coated with epoxy up to 3 mm from the tip, the teeth in this part of the microelectrode that couples with the comb-drive is no longer functional. As a result, the overall translational capability of our electrostatic microdrive after insulation is now reduced to approximately 2 mm. The electrostatic microactuator is able to generate force in the order of micro-Newtons and has been tested for actuation within Jell-O[®] as well as rat brains using the design with two torque amplification stages of 12 : 1. However, the force is not sufficient to penetrate the dura mater in rats. We have found that the power dissipation in the electrostatic microactuator is negligible. While input voltages are in the order of 90 V to drive a bank of capacitors in the comb-drive, the current drawn is in the order of a few nanoamperes. The power dissipation is, therefore, in the order of fraction of microwatts. The speed of actuation is largely dependent on the frequency of the sinusoidal waveforms used to activate the electrostatic microactuators. Each complete cycle of the rectified, optimized sinusoidal waveform result in one rotation of the pin-joint.

We have tested the electrostatic microactuators with sinusoidal frequencies up to 1000 Hz. However, higher frequencies in the order of several kilohertz are routinely used during bench testing of the microactuators. Two of our microactuators also failed due to stiction between the fingers of the bank of parallel plate capacitors in the comb-drive upon activation. Typical impedance spectra (both magnitude and phase) of the polysilicon microelectrode measured while the microactuator was deactivated are shown in Fig. 9. The magnitude of the impedance at 1 kHz was 1.6 M Ω . While the electrical recording capabilities of isolated polysilicon microelectrodes have been documented earlier [26], multiunit activity was also obtained with the polysilicon microelectrodes coupled to the electrostatic microactuator as shown in Fig. 10. The spike amplitudes were in the range of 100–500 μV and the RMS value of the noise was approximately 8–10 μV . The average SNR (over all the units) was 18–20 dB. The electrostatic microactuator was deactivated during the time of recording to avoid electrical artifacts from the activating waveforms.

IV. Discussions

The overall translational capability of our electrostatic microdrive after insulation is now reduced to approximately 2 mm. In the future, the process for insulating the microelectrodes can be integrated seamlessly with the SUMMIT VTM technology that will preserve the teeth along the sides of the microelectrode shank and will help us retain the full 5-mm-translational capability of our microdrive. However, if the object of microactuation *in vivo* is to reposition the microelectrode in order to identify another neuron in the vicinity of the one with which electrical contact has been lost, then it is our experience that typically a few tens or at most hundreds of microns of actuation is sufficient to re-establish electrical contact with another neuron. Therefore, with its present translational capabilities, our electrostatic microdrive can potentially help prolong electrical monitoring of single neurons in chronic experiments.

In an earlier study, we had reported that the approximate force required to penetrate the rat dura with polysilicon microelectrodes that were not tapered (approx. $50 \times 4 \mu\text{m}$ in cross-

section) to be in the order of 1 mN [26] when it was moved in increments of 10 μm . Therefore, the force generated by the electrostatic microactuator (in the order of micro-Newtons) is probably insufficient to penetrate the dura mater without torque amplification stages. However, it remains to be tested if the two 12 : 1 torque amplification stages used here will generate sufficient force to penetrate the dura. There is no heating of the microactuator engine in the electrostatic microactuators. This implies that several microactuator chips can likely be stacked close to each other for a multiple array electrode configuration without the need for specialized heat sinks. One of the critical issues for successful actuation of the electrostatic microactuators is precise synchronization of activating waveforms with a 90° phase difference between the *x*-drive and the *y*-drive. Failure due to stiction between the fingers of the bank of parallel plate capacitors in the comb-drive upon activation could likely be due to unbalanced forces resulting from a fault in the position of the interdigitated fingers. During *in vivo* testing in two of our acute experiments, blood clots had formed around the microelectrode due to bleeding from internal microvasculature. The adhesion of the blood clots to the microelectrode was strong enough to break the microelectrodes in between along the length of the shank when we attempted to gently pull it from the brain in 10 μm increments. Breakage of microelectrodes in this fashion could be a serious failure mode in long-term implants. Bioactive coatings that reduce cellular adhesion [27] on the surface of the microelectrodes can help in reducing tensile stress build-up and breakage while pulling the microelectrode out of the brain. Another strategy that is likely to help under the above conditions is to first move the microelectrode into the brain for approximately 200 μm in steps of 10 μm or less to relieve any residual adhesion before reversing the microactuators and pulling the microelectrode out. Since the polysilicon microelectrodes bend quite easily, stress required to break the polysilicon microelectrodes is very high under compressive entry into the brain compared to the tensile stress required for plastic deformation while extracting the microelectrode. The polysilicon microelectrodes are very thin (approx. $50 \times 4 \mu\text{m}$) compared to the cross section of other microfabricated silicon based microelectrodes [28]. The mechanical properties of the polysilicon shanks without taper have been discussed in our earlier study [26].

In addition, we have found that the electrostatic microactuators and the associated gear trains are very sensitive to the alignment of gears. We found a few instances when the microelectrode stalled likely due to a misalignment of gears. In some of these cases, we got the microactuators to work with a gentle push along the microelectrode shank. This will be a critical issue in long-term experiments. Behaving rodents go through a variety of movement routines like grooming that will likely be extremely stressful on the mechanical stability of the microactuated microelectrode. While the use of epoxy as an insulator has worked very well in our acute experiments, proven insulators with better stability in biological environment like parylene-C or silicon nitride will be a better choice of insulator under chronic conditions. Silicon nitride can be readily integrated into the SUMMiT V™ process and, therefore, has the advantage that it will likely allow us to retain the overall translational capability of 5 mm for the microelectrode even after insulation. Clearly, insulation of the polysilicon microelectrode needs to be optimized for stability and biocompatibility for such chronic experiments and packaging issues will have to be carefully thought through as well.

Unlike fixed microelectrode arrays, complete hermetic packaging of the electrostatic microdrive will not be possible since the microelectrode will be moving in and out of the chip even after implantation. Therefore, moisture will also be an important factor that determines the functionality of the device in long-term *in vivo* brain experiments. Hydrophobic coatings on the packaging and the microelectrode itself will offer some protection against entry of fluids into the micromachinery. However, a rigorous testing of the electrostatic microactuator under moist conditions will be a necessary first-step. Therefore, several challenges remain before this device can be used in chronic or long-term experiments.

V. Conclusion

In conclusion, we have successfully demonstrated in acute experiments an electrostatic microactuator technology that can be used to reposition microelectrodes in the rodent brain tissue. The polysilicon microelectrodes were capable of monitoring single neuronal electrical activity when the microactuators were deactivated. This is the first successful demonstration of a microactuated microelectrode technology using batch fabrication MEMS procedures like the SUMMiT V™ process. Our study here demonstrates that the microactuated microelectrode will enable precise positioning and recording from single neurons in acute experiments obviating the need for any manual operation of the stereotactic controls. However, the more significant application of this device lies in long-term positioning and recording applications *in vivo*. In such applications, the device will allow for precise repositioning of the microelectrodes in the event of loss of electrical contact. However, several developmental issues related to insulation, packaging and electrical interconnects will need to be optimized before this technology can be applied to long-term recordings.

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Aaron Gilletti received the B.S. and M.S. degree in the Harrington Department of Bioengineering, Arizona State University, Tempe, in 2001 and 2003, respectively. His thesis work involved BioMEMS for neuro-electrophysiology signal acquisition and the quantification of tissue micromotion. His research interests include BioMEMS technologies for neural interfaces.

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Fig. 1.

(a) Principle of electrostatic actuation between parallel plates (comb structures). (b) Schematic of the basic electrostatic comb-drive actuator subassembly. Typically, two of these units are cascaded to form a single actuator. Schematic of two actuators (x - and y -drives) with their associated link arms (c) at rest (or fabricated) positions (d) acting synchronously to create a rotational motion at the end. (e) A sketch of the entire actuation assembly showing several banks of parallel plate capacitors constituting the x - and y -drives. The x - and y -drives are coupled to the output gear through a pin-joint.

Fig. 2. Schematic of the SUMMiT V™ process indicating the different layers of polysilicon available for fabricating mechanical and electrical components.

Fig. 3.

(a) Design of the microactuator coupled with the microelectrode (horizontal shuttle at the bottom of the figure approximately 5 mm long) through transmission gears. The x -drive and y -drives are shown 90° apart. (b) Design of the microelectrode showing the laminated layers of polysilicon that make up the microelectrode and also the guide rails that go over the microelectrode. (c) An illustration (not drawn to scale) of how the microactuated microelectrode positioned on top of the cortical surface will be eventually used *in vivo*. The electrostatic microactuator can drive the microelectrode into the brain up to a maximum distance of 5 mm in this design, which is sufficient to probe the deep cortical, subcortical and hippocampal structures in the rodent.

Fig. 4. Micrographs showing two different design variations attempted. (a) The microelectrode is driven by a 2-stage transmission (12 : 1 gear ratio at each stage) between the microelectrode and the microengine. (b) The microengine is directly connected to the microelectrode.

Fig. 5. Micrographs of the microelectrode showing that in both designs, there are guides that go over the microelectrode and a pin-in groove alignment structure.

Fig. 6. Micrograph of the microelectrode *after fabrication*. The microelectrode (width 50 μm, length of taper 100 μm) is shown along the right edge of the chip. Teeth microfabricated on one edge of the microelectrode enable coupling between the microelectrode and the transmission gears.

Fig. 7.

(a) A picture of the polysilicon microelectrode extended out of the edge of the chip. Electrical activity is tapped out by a silver wire bonded to the microelectrode from the side. (b) A microactuated microelectrode chip shown alongside a penny for size comparison.

Fig. 8.

Waveforms used for activating the microactuators during three different phases of motion—during startup from rest, actuation by one-step after startup and subsequently bringing the actuator to rest. The *up* and *down* waveforms power the *x*-drive and the *left* and *right* waveforms power the *y*-drive. The waveforms that power the *x*-drive and those that power the *y*-drive have a 90° phase difference. In addition, the waveforms corresponding to the *up* and *down* actuation sites are two halves of a rectified, optimized sinusoid. The same is true of the waveforms corresponding to the *left* and *right* actuation sites. The velocity and step-size can be altered by suitably changing the frequency and number of cycles (or fraction thereof) of the actuating waveforms.

Fig. 9.
Magnitude and phase of the impedance spectrum of the polysilicon microelectrode insulated with epoxy.

Fig. 10.
Multiunit data from the polysilicon microelectrode.