

MSEC2026-177108

DEVELOPMENT AND INVESTIGATION OF A PIEZOELECTRIC INCHWORM ROBOT FOR NEURAL INTERFACE IMPLANTATIONS

Kevin Lat
Department of
Mechanical and
Industrial
Engineering,
University of
Massachusetts
Lowell
Lowell, MA

Xiaowen Chen
Department of Electrical
and Computer
Engineering, University
of Massachusetts
Lowell
Lowell, MA

Noah Jafferis
Department of Electrical
and Computer
Engineering, University
of Massachusetts Lowell
Lowell, MA

Lei Chen
Department of
Mechanical and
Industrial Engineering,
University of
Massachusetts Lowell
Lowell, MA

ABSTRACT

Large-scale chronic recording across the brain with minimal disruption will be critical for understanding the multi-region neural dynamics. But the possibility of such experiments is limited by buckling of the electrodes upon insertion. The least-damaging microelectrode arrays (MEAs) would be preferred for broad chronic recording but buckle against brain membranes during insertion, so larger more damaging electrodes/shuttles are usually used. To address this challenge, in this study, a piezoelectric inchworm insertion robot was developed to enable currently impractical brain implantation of miniaturized flexible MEAs through the dura and pia mater without buckling in a fast and extensible machine-controlled manner. The inchworm inserter robot conducts iterative grip-feed-release inchworm motion holding the electrode body. This reduces buckling by increasing lateral extracranial support along the electrode during implantation and minimizing unsupported length (maximizing the critical buckling load). The prototyped inchworm inserter was composed of three custom piezoelectric unimorph “re-curve” actuators: two horizontal grippers alternately gripping the implant and an extender that moves the grippers vertically. The assembly is of miniaturized size of $14 \times 5.8 \times 3.6$ mm and 250 mg in weight. Custom designed piezoelectric ‘re-curve’ actuators together with adjustable gripping plates were used to produce up to 80 μ m of purely linear motion with minimal bending, up to 950 mN of gripping/insertion force, and up to 200 Hz of bandwidth, enabling up to 14 mm/s precisely controlled insertion speed and iterative increments as little as 10 μ m. These specs were all further tunable by varying the actuator design and power input. Benchtop tests were conducted against a multi-layer (dura-pia-

brain tissue) rodent brain-mimicking phantom. The tests with the inchworm insertion robot demonstrated successful buckling-free penetration of dura and pia membrane phantoms with 12 and 25 μ m diameter tungsten microwires and 15 μ m-thick single-, dual-, and four-shank silicon probes. In comparison, under conventional stereotaxic insertions, these devices were limited to pia-only insertion and would buckle against the dura mater. The inchworm insertion motion also reduced the membrane rupture force and dimpling depth compared to static stereotaxic insertion methods, with optimal performance observed around medium 75 Hz operation frequency for 25 μ m diameter tungsten microwires. These results highlighted the great potential of the piezo inchworm insertion robot to achieve broadly applicable improvements in brain surgery methodology by enabling thick membrane penetration by miniaturized MEAs and machine-controlled automatic buckling-free insertion with precise control of the speed and depth.

Keywords: Neural interface, inchworm motion, piezoelectric actuator, insertion robot, buckling prevention.

NOMENCLATURE

F	Insertion force
P	Critical buckling load
L_U	Electrode unsupported length
E	Elastic modulus
I	Moment of inertia
n	Buckling mode constant
L_A	Length of piezoelectric actuator
W	Width of piezoelectric actuator
T	Thickness of piezoelectric actuator
D_f	Free deflection of piezo actuator

D_b	Actuator deflection due to initial gap to sensor
$F_{b, meas}$	Measured blocked force of actuator
F_b	Calculated blocked force of actuator

1. INTRODUCTION

The brain functions through the firing of neurons via an electrical signal, known as an action potential [1]. To fully understand the brain functioning mechanism, it is critical to achieve high temporal and spatial resolution recording of such single-neuron action potentials across brain regions and over long periods of time [2, 3]. Implantable microelectrode arrays, placing recording electrodes right next to the firing neurons, are an inexpensive and well-developed tool that has significant advantages of recording speed and specificity to achieve single-neuron activity recordings [4]. To enable large-scale high channel count recording and to ensure chronic signal quality without triggering the immune response, smaller and more flexible microelectrodes with properties similar to the soft brain tissue are preferred [5-8]. Recent development of such small and flexible devices include microwire arrays made of miniaturized 5 - 10 μm diameter carbon fibers [9-12] and flexible substrate planar neural probes made of Parylene-C, Polyimide, PDMS, SU-8, etc., [13-16]. However, these least-damaging electrodes are also the most difficult to use with current manual stereotaxic surgical methods, facing major barrier in electrode buckling when pushed longitudinally against the protective brain membranes (e.g., the dura and pia maters).

The microelectrode implantation into the brain through the brain membranes usually ends up with two results (as shown in Fig. 1): (1) buckling against the membrane, or (2) successful membrane penetration without buckling. Microelectrode buckling happens when the insertion force F exceeds the electrode's critical buckling load P . The critical buckling load P is commonly calculated based on Euler's column equation [17]:

$$P = \frac{n\pi^2 EI}{L_U^2} \quad (1)$$

where L_U is the electrode unsupported length, E is the elastic modulus, I is the moment of inertia, and n is the buckling mode constant ($n = 0.25$ for a fixed-free end condition and $n = 2$ for a fixed-pinned end condition). When electrode buckling happens, it can be easily identified via visual observation and a sudden drop on the insertion force profile. The insertion process will be halted to prevent electrode damage and/or significantly tilted insertion and resulting tissue damage. On the other hand, if the electrode critical buckling load P is bigger than the brain membrane rupture force, the electrode will penetrate the membrane buckling-free and insert into the brain. Upon membrane rupture, since further resistance due to electrode-tissue friction is much smaller than the further increased critical buckling load (due to smaller unsupported length), buckling change is negligible. To prevent electrode buckling, prior studies have either (1) decreased insertion force F , or (2) increased the critical buckling load P . While insertion force reduction efforts through tip sharpening [18-22] and vibration-assisted insertion [23, 24] showed some promising results against the thin pia mater penetration, penetration of the thin and tough dura mater

has mainly remained infeasible, even for smaller rodent models [25]. Research on increasing the buckling load P has 1) switched to greater modulus and/or larger size electrodes, 2) used permanent supports to strengthen the structure (e.g., silicon and diamond shanks) [23, 26], 3) utilized large retractable shuttles or stiff materials like metal needles [27, 28], 4) bundled minimally invasive electrodes together, and 5) used temporary support materials, like bio-dissolvable epoxy, dissolved away during [26] or after implantation [29]. These methods either caused more damage to the brain or added complexity to the already labor-intensive surgery.

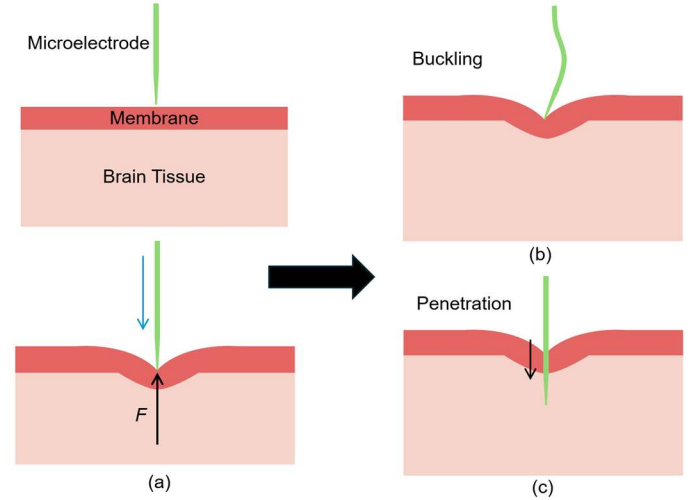


FIGURE 1: MICROELECTRODE INSERTION AGAINST THE BRAIN MEMBRANES: (A) RESISTANCE FROM THE DIMPLED MEMBRANE BEFORE RUPTURE, LEADING TO TWO POTENTIAL RESULTS OF (B) BUCKLING AGAINST THE MEMBRANE OR (C) SUCCESSFUL MEMBRANE PENETRATION WITHOUT BUCKLING.

More recently, studies have been conducted to minimize the buckling chance via reducing the unsupported length L of the electrode. Chen et al. [30] developed a 3D printed skull cap with precision molded guides to provide full support along the electrode insertion path. A pair of wire grippers (e.g., tweezers) were used to incrementally insert the electrode into the brain. The experiment showed promising results enabling previously impossible rat dura penetration for 25 μm tungsten microwires and 7 μm carbon fibers. Based on such a concept, a DC-motor-driven, synchronized dual cam controlled incremental inchworm insertion machine was designed and prototyped, which enabled straight insertion all the way through a 22 mm thick PVC phantom as compared to wire buckling under the conventional back-end support insertion approach [17]. Despite the promising developments and test results, the motor and cam wheel driven inchworm insertion device was of bulky size, preventing them from handling miniaturized electrodes in tight space like within a rodent skull area. Similarly, Smith et al. developed an inchworm mechanism system with melting phase-change materials and piezo actuator motion to insert microwire and wire tetrode arrays into different brain depths [31]. The phase-change

material-based gripping mechanism limited the device to be applicable to only cylindrical shaped electrodes. Due to the slow melting nature of phase-change materials, such inchworm mechanism inserter might further slow down the already labor-intensive and hours-long electrode implantation surgery. Also, while the piezoelectric block stack actuator enabled precise control of the insertion motion, its large size required the implantation electrode to be centimeter long, which might not be practical for most brain recording devices.

To address the size and insertion speed issue of the current devices and to further enhance the inchworm mechanism towards automated brain electrode insertion, in this study we designed and fabricated a piezoelectric inchworm robot to automate the grip-insert-release motion using three custom piezo unimorph recurve actuators, making it of miniaturized size and light weight and thus applicable to high-density neurosurgical implantations in tight spaces. The developed inchworm robot was also tested against a multi-layer brain-mimicking phantom to validate its buckling prevention and automated insertion capabilities.

2. MATERIALS AND METHODS

This section describes the inchworm motion mechanism, our design and fabrication of the inchworm robot, and proof-of-concept test setup and data processing methods.

2.1 Inchworm Robot Insertion Mechanism and Design

The incremental insertion cycle of an inchworm insertion mechanism is as shown in Fig. 2. To realize such grip-insert-release mechanism, at least three actuators were needed as (1) upper and (2) lower pairs of electrode grippers and (3) a vertical extender for the longitudinal insertion motion. Correspondingly, as in Fig. 2(a), we had three piezoelectric actuators included in the inchworm robot design, with neutral position of the horizontal actuators corresponding to open status of the upper and lower grippers while the vertical actuator extended to activate the insertion motion.

Before each insertion, the microelectrode slid into the grippers as initial alignment, as in Fig. 2(a). Then, the lower gripper was closed for the initial grip (Fig. 2(b)), followed by activated vertical extender to raise the upper gripper (Fig. 2(c)). Next, the upper gripper was activated to hold the electrode (Fig. 2(d)) before release of the bottom gripper (Fig. 2(e)). At this time, the inchworm was ready for the insertion motion, with the distance between the top gripper and the brain surface as the maximum unsupported length, instead of the whole electrode length as unsupported in conventional stereotaxic insertions. The incremental insertion of the electrode in this cycle (Fig. 2(f)) was conducted by retraction of the vertical extender piezoelectric actuator. After insertion, the lower gripper was closed to hold the electrode (Fig. 2(g)) before the upper gripper was released (Fig. 1(b)). At this point, the inchworm actuator had returned to the original position, ready for the next cyclic motion to start.

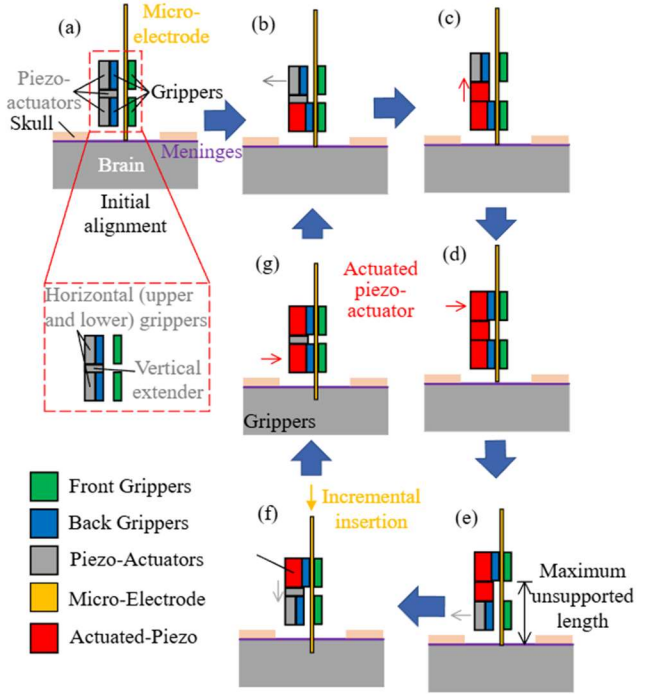


FIGURE 2: STEPS INVOLVED IN AN INCHWORM MOTION INCREMENTAL INSERTION CYCLE.

For the two grippers alternatively gripping the electrode and the vertical extender that moved the electrode along the insertion direction, they were required to provide precise motion with minimal bending, as the electrodes (especially silicon-based neural probes) were fragile. To achieve that, piezoelectric “re-curve” unimorph actuators [32] (Fig. 3) was used in this study. Such actuators could produce 10s - 100s μm of purely linear motion with minimal bending, 100s of mN force range (larger than our published rat dura penetration force at 10s of mN for 100 μm diameter microwires [33]), and 100s of Hz bandwidth. More importantly, such motion and force range were tunable by modifying the actuator design. The “re-curve” unimorph actuator had deflection output scaling with L_A^2/T and the output force range was proportional to WT^2/L , where L_A , T , and W were length, thickness, and width of the custom piezoelectric actuator designs. Thus, while the inchworm robot presented in this study was mainly designed towards microelectrode insertion into brains, the piezoelectric actuator-based inchworm insertion robot concept could be widely applied to other implantation cases with buckling prevention needs, via customizing the piezoelectric actuator dimensions accordingly.

In this study, based on preliminary insertion tests results, it was concluded that actuated gripper motion of 60 μm was appropriate for microelectrode alignment. Based on that, the actuator length was chosen to be 14 mm (with 10 mm active length) to provide such 60 μm opening at 200 V using the thinnest easily available PZT-5H sheets (135 μm thick). The width of 2 mm was then chosen based on the maximum expected

force requirements for the electrode insertion (10s of mN), together with a safety margin to account for the coefficient of friction and actuator performance variation, as well as practical considerations such as having room for the electrical connections and for the gripper latch plates' support rods.

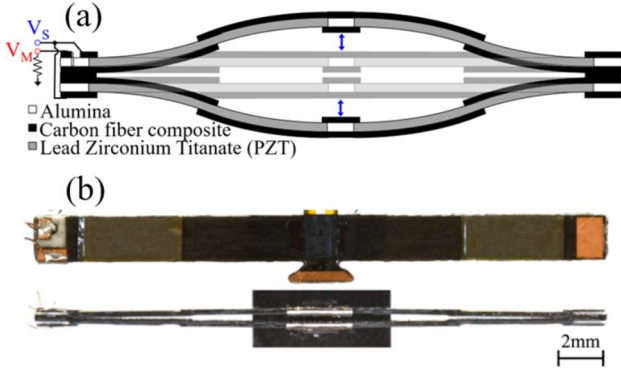


FIGURE 3: A SAMPLE PIEZO UNIMORPH RECURVE ACTUATOR (25 MM BY 2 MM PROTOTYPE SHOWN): (A) CROSS-SECTIONAL DIAGRAM AND (B) PHOTO OF PHYSICAL PIECE.

2.2 Fabrication of Piezoelectric Actuators

Fabrication of the three piezoelectric “re-curve” unimorph actuators followed the bimorph fabrication process as elaborated in our previous study [34]. The fabrication process started by laser-cutting of all material layers placed on gelpak, followed by ultrasonic isopropyl alcohol (IPA) cleaning of lead zirconate titanate (PZT) and copper layers to ensure strong bonding. A dummy Kapton layout was used to pre-mold pacopad for uniform pressure during curing. Uncured carbon fiber (CF) sheets were then inspected, cut, and stored before assembling half-stacks, which underwent heat-press lamination. Pre-release raster and cuts defined electrical traces and relieve stress, while epoxy overflow in holes was cleared manually. To bond the two half-stacks together, another uncured CF layer was used in a second heat-press lamination process. Then, final release cuts separated the actuators (Figs. 4(a) and (b)), which were then cleaned thoroughly with IPA. Electrical insulation was applied via cyanoacrylate glue along the cutting edges of the actuators (or optionally coat the entire actuators with Parylene), and conductive epoxy connected copper traces and vias. Spacer layers of Kapton were added for gripper alignment, and actuators were wired for single-actuator performance tests to identify any fabrication defects (Fig. 4(c)). Free deflection was measured under an optical microscope, and the blocked force was measured against a force sensor (Nano 17 by ATI Industrial Automation, Apex, NC). For the first inchworm prototype tested in this study, the three lowest performing actuators were used to provide a margin of error for insertion capability results.

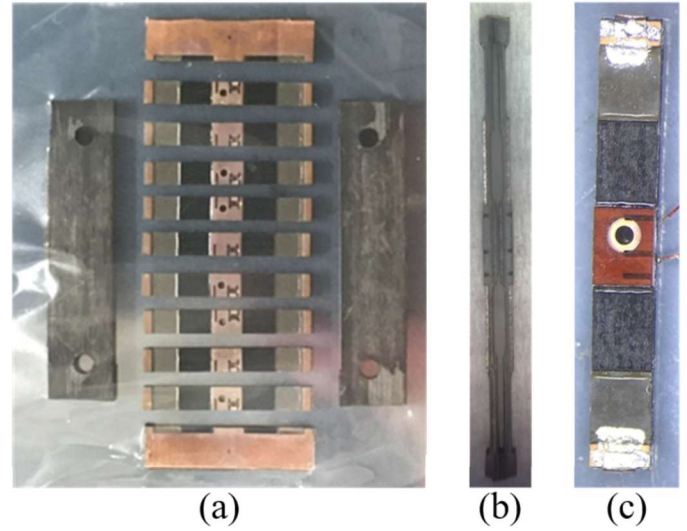


FIGURE 4: FABRICATION OF THE PIEZO UNIMORPH RECURVE ACTUATORS: (A) SEPARATED ACTUATORS FROM THE TWO-STEP HEAT-PRESS CURED STACK, (B) EDGE-VIEW OF AN ACTUATOR AFTER RELEASE, AND (C) KAPTON SPACERS ADDED USING REMOVABLE TAPE TO ALLOW FOR FORCE MEASUREMENTS ON THE INDIVIDUAL ACTUATOR.

2.3 Inchworm Robot Assembly

With the three fabricated single piezo unimorph actuators, the inchworm insertion robot assembly process began by mechanically joining the two horizontal gripper actuators to the extender actuator using FR4 ribs fitted into angled trenches and glued while maintaining co-planarity. Electrical connections were established through small wires linking actuator traces to laser-etched copper-clad FR4 plates, which were then tinned and mounted on the ribs (Fig. 5(b)). A mounting fixture was created using precision-aligned nuts, FR4, and Kapton spacers, and reinforced braces, with a custom carbon-fiber screwdriver fabricated for handling tiny screws (Figs. 5(a) and (b)). Each horizontal actuator conducted the electrode gripping function via a pair of latch plates (160 μm thick each made of carbon fiber sheets). The inner latch plates (blue gripper as in Fig. 2) were attached to Kapton spacer layers, while outer latch plates (green gripper part as in Fig. 2) were mounted on carbon-fiber rods (500 μm diameter) passing through actuator holes and secured with removable UV-cure adhesive (Figs. 5(c) and (d)). Mylar strips were added around the rods to prevent electrical shortage, and carbon-fiber washers were used to strengthen rod-to-plate joints. Additional CF reinforcement pieces were glued to FR4 for rigidity, and wiring was completed with short and long American wire gauge (AWG) wires arranged to minimize stiffness during operation (Fig. 5(c)). To improve alignment and stiffness, extra CF latch plates were added first and then laser-cut apart, and removable adhesives were applied for modularity. Finally, CF supports were added to the upper gripper rod to prevent tilting under load, ensuring structural stability and adjustability for the prototyped inchworm insertion robot. The final inchworm insertion robot assembly is as shown in Figs. 5(e) and (f). To hold such assembled inchworm device onto a test fixture in this

study or similarly onto a stereotaxic device during further animal tests, a 3D printed holder was custom designed, fabricated, and installed through the precision aligned nuts in the assembly, as shown in Figs. 5 (c) and (d).

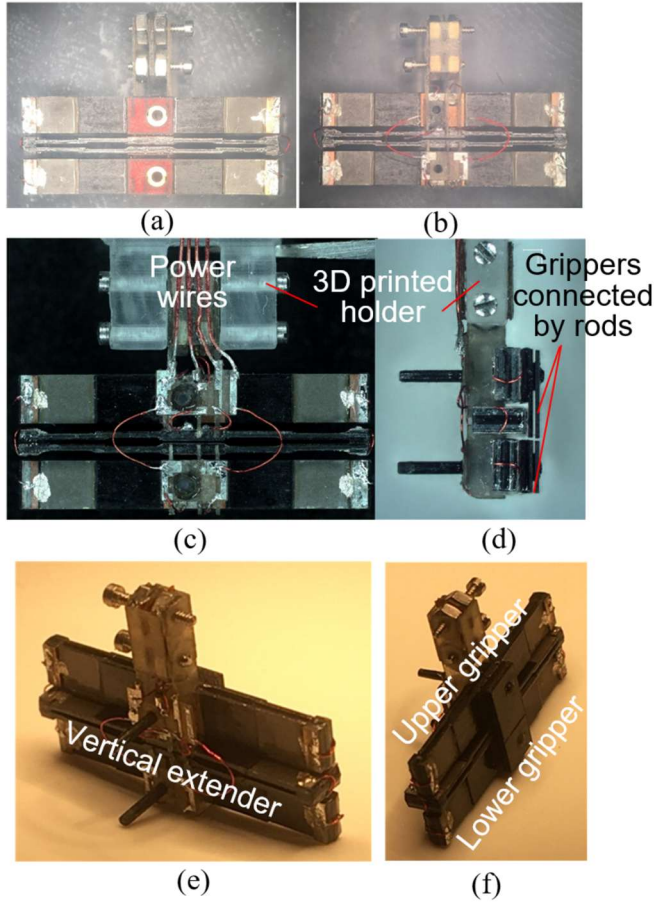


FIGURE 5: ASSEMBLY OF THE FABRICATED PIEZOELECTRIC ACTUATORS AND CUSTOM GRIPPERS FOR THE INCHWORM ROBOT: (A) THREE ACTUATORS ARE AFFIXED TOGETHER WITH FR4 RIBS TO FORM THE INCHWORM CONFIGURATION, (B) CARBON FIBER PLATES ADDED WITH WIRES SOLDERED, (C) BACK AND (D) SIDE VIEW OF ASSEMBLED INCHWORM WITH MAIN EXTERNAL POWER WIRES SOLDERED AND GRIPPERS WITH CONNECTING RODS INSTALLED, AND (E) AND (F) ASSEMBLED INCHWORM ROBOT WITH 3D-PRINTED TEST FIXTURE FOR HOLDING THE INCHWORM.

2.4 Inchworm Robot Insertion Tests

Proof-of-concept inchworm insertion tests were conducted with the assembled inchworm robot using the setup as shown in Fig. 6(a). The whole setup was placed on an optical board with vibration isolating feet. The assembled inchworm insertion robot was held by an optical post, which could move within the horizontal plane for alignment through two manual linear stages.

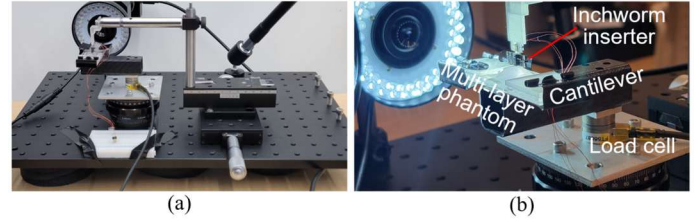


FIGURE 6: EXPERIMENTAL SETUP FOR THE INCHWORM INSERTION TESTS: (A) SETUP OVERVIEW AND (B) CLOSE-UP VIEW OF THE KEY INSERTION AREA AT THE INCHWORM-PHANTOM INTERFACE.

The inchworm insertion target was a multi-layer brain-mimicking phantom [35]. As elaborated in Fig. 7, the multi-layer phantom was composed of a stretched PVC layer to mimic the dura mater, a 1.01% weight/volume ratio Agarose layer as the pia mater, and a 0.5% weight/volume ratio Agarose layer to replicate the soft brain tissue. The properties of this multi-layer phantom was calibrated and validated against in-vivo rat brain insertion tests [33] and showed comparable rupture force and dimpling depth output under both pia-only and dura-pia insertion conditions of various sized microwires and silicon-based electrodes, similar to the ones used in this study.

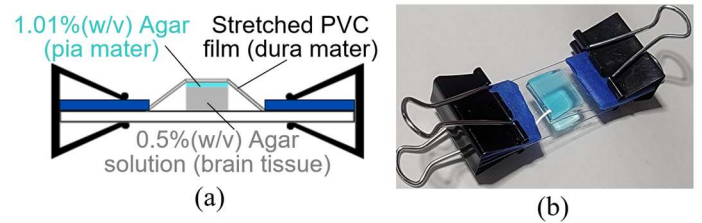


FIGURE 7: MULTI-LAYER DURA-PIA-BRAIN MIMICKING PHANTOM DUPLICATING IN-VIVO RAT BRAIN PROPERTIES DURING INSERTION: (A) DIAGRAM OF DIFFERENT PHANTOM LAYERS AND (B) PHOTO OF THE ASSEMBLED MULTI-LAYER PHANTOM.

During the insertion tests, as shown in Fig. 6(b), the multi-layer brain-mimicking phantom was placed on top of a load cell (Nano 17 by ATI Industrial Automation, Apex, NC) for the insertion force measurement. Our preliminary insertion trials showed excessive noise levels induced by environmental vibrations and air flows, inducing a low signal-to-noise ratio. Thus, a cantilever beam structure, as shown in Fig. 6(b), was fabricated out of a carbon fiber square tube and converted the longitudinal insertion force measurement into bending torque measurement by the load cell, dramatically reducing the impact from environmental noises. Before each test, the electrode to be inserted was first slid into the horizontal gripper plates and held in place. Then, the manual lab jack at the bottom of the load cell was used to vertically align the multi-layer phantom top surface closely with the inchworm-held microelectrode tip (within mm-level gap). Next, the inchworm robot was actuated at different frequency levels (0.5, 1, 2, 5, 10, 15, 25, 50, 75, 100, 150, and 200 Hz tested) to conduct the incremental insertion tests. Under each frequency, at least three trials were conducted and no

buckling was observed under these frequencies. During each test, the load cell recorded the bending moment at 1000 Hz sampling rate up until at least 1 mm tip penetration into the brain phantom to ensure accurate rupture force and dimpling depth identification. Insertion tests were conducted at least 0.5 mm away from previous phantom rupture points to minimize the impact of pre-ruptured membrane phantoms.

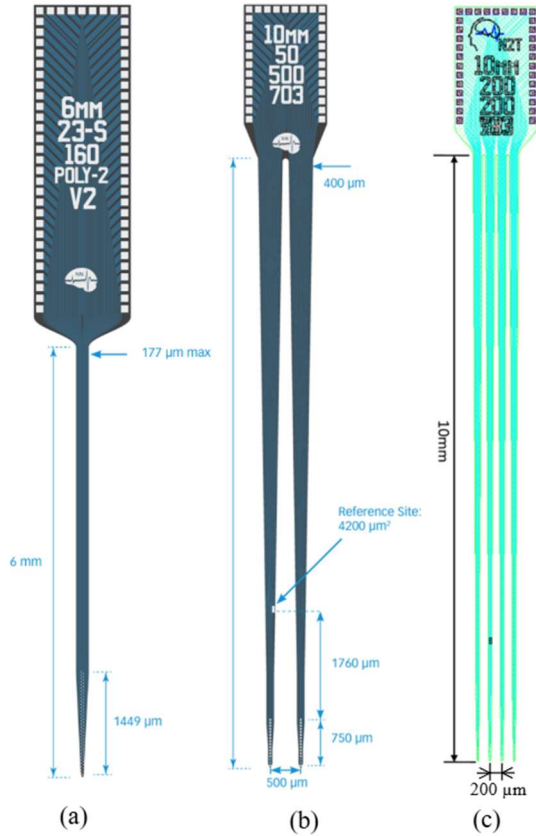


FIGURE 8: DIMENSIONS OF THE PLANAR NEURAL PROBES (ALL FROM NEURONEXUS WITH 15 μM THICKNESS) USED IN THE INSERTION TESTS: (A) SINGLE SHANK A1X64-EDGE-6MM-50-177 PROBE, (B) DUAL SHANK A2X16-10MM-50-500-177 PROBE, AND (C) QUADRO SHANK A4X8-10MM-200-200-413 PROBE.

To ensure the motion synchronization of the three piezoelectric actuators, each actuator was independently controlled with voltage output from a data acquisition (DAQ) device and high-voltage amplifier. The DAQ had up to 1 MHz update rates (resulting in $\sim 1 \mu\text{s}$ delays between the different output channels), and the amplifier was rated to supply the currents necessary to run the actuators over the frequency range tested with negligible timing errors. The control voltages were chosen so that there was no time at which both grippers were open, and to ensure that the extender did not move during the brief periods that both grippers were closed. Such synchronization was confirmed via high-speed camera recording of inchworm insertion tests.

In this study, five different types of electrodes were tested: (1) 25 μm diameter bare tungsten microwires (Midwest Tungsten Service, Willowbrook, IL), which were tested under all twelve inchworm frequencies with the minimal rupture force (to be shown in the Results section) found at 75 Hz and thus, used for all other four types of electrodes, (2) 12 μm diameter insulated tungsten microwires (California Fine Wire Co., Grover Beach, CA), (3) 15 μm thick single-shank silicon-based planar neural probe (Fig. 8(a)), (4) 15 μm thick dual-shank probe (Fig. 8(b)), and (5) 15 μm thick quad-shank probe (Fig. 8(c)). Each silicon probe, with detailed dimensions shown in Fig. 8, was acquired from NeuroNexus Technologies, Inc. as dummy duplicates, which represented same mechanical properties as functional probes for the insertion tests in this study. These were tested under optimal 75 Hz frequency in this study. These five electrodes were chosen as common surgically used ones that were prone to buckling, especially against the rodent dura mater, which were conventionally treated as impossible to penetrate.

2.5 Insertion Profile Analysis

Sample insertion force against time profile is as shown in Fig. 9, extracted from a 25 Hz inchworm insertion test of a 25 μm diameter bare tungsten microwire against the multi-layer brain-mimicking phantom. Such a profile was obtained by converting the load cell recorded bending moment – time dataset into force based on the cantilever beam offset distance. In this study, electrode buckling was monitored via the real-time camera as well as the insertion force profile. The membrane penetration was confirmed from two aspects (as shown in Fig. 10): (1) sudden drop of insertion force profile together with visual confirmation of dimpling reduction and electrode tip penetration and (2) negative insertion force readings during retraction process of the electrode.

From the insertion force profile, a peak force level followed by a sudden insertion force drop identified the membrane rupture point. The corresponding force level was the membrane rupture force, which was important for electrode buckling if it went beyond the electrode critical buckling load. Based on the time of rupture point and the inchworm feeding speed / number of cycles conducted, the insertion force profile also provided an estimate of the membrane dimpling depth at rupture, commonly used by surgeons to determine insertion resistance during animal surgeries when real-time force measurement is difficult. Besides those membrane-rupture-related data, friction during further insertion as well as retraction forces from the brain (not shown in Fig. 9) could also be extracted from the insertion force profile if of specific interest for the study.

Note that for higher frequency inchworm tests, significant number of fluctuations and random noise showed up in the insertion force profile. To smooth out the insertion force profile, a simple moving average filter (11-point average) was applied to help identify the rupture point and thus the corresponding membrane rupture force and dimpling depth at rupture.

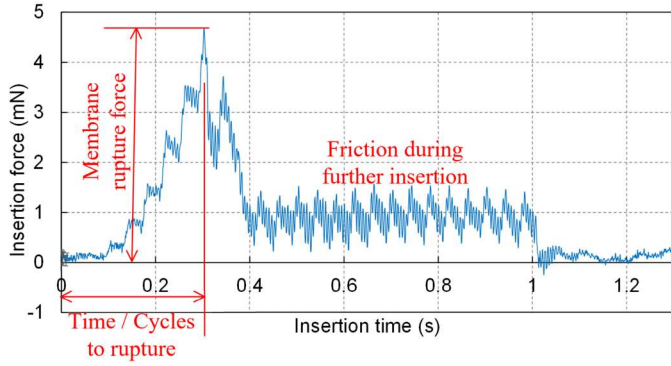


FIGURE 9: SAMPLE INSERTION FORCE PROFILE FROM A 25 HZ INCHWORM ROBOT INSERTION TEST OF A 25 μm DIAMETER BARE TUNGSTEN MICROWIRE AND KEY INSERTION OUTCOMES EXTRACTED FROM THE FORCE PROFILE.

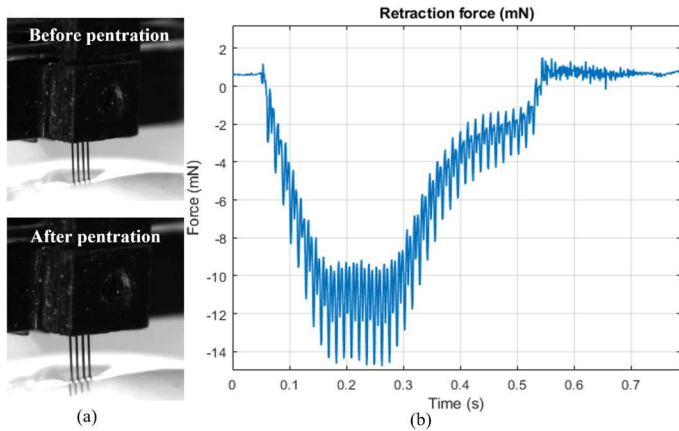


FIGURE 10: METHODS FOR MEMBRANE PENETRATION CONFIRMATION: (A) REAL-TIME IMAGING FOR VISUAL CONFIRMATION OF MEMBRANE PENETRATION AND (B) SAMPLE RETRACTION FORCE PROFILE FROM A 75 HZ INCHWORM ROBOT RETRACTION TEST OF QUADRO SHANK 18 μm NEURONEXUS PROBE.

3. RESULTS AND DISCUSSION

3.1 Inchworm Insertion Robot Characterization

Following the fabrication and assembly steps elaborated before, the finally achieved piezoelectric inchworm insertion robot is as shown in Fig. 11, with its size compared with a U.S. quarter dollar coin. Characterization tests showed that the assembled inchworm device was of $14 \times 5.8 \times 3.6$ mm size in total and of 250 mg weight, which was miniaturized for high-density electrode insertions within tight skull-top space.

The assembled inchworm was able to provide up to 80 μm linear motion range at 250 V for gripping and insertion motion via the three independently controlled actuators. The two horizontal grippers, with temporary spacers that added a custom gap, were able to provide a 40 μm open gap for initial alignment of the 12 – 25 μm thickness electrodes used in this study. The iterative motion increments could be controlled to as little as 10 μm , enabling high-precision control of the insertion depth. But

the most consistent insertion distance was found to be 67 μm at 200 V, the main setting used for all trials in the results section. Once a voltage was applied a high-definition camera and motion tracking was used to measure the distance the actuator produced to ensure accuracy.

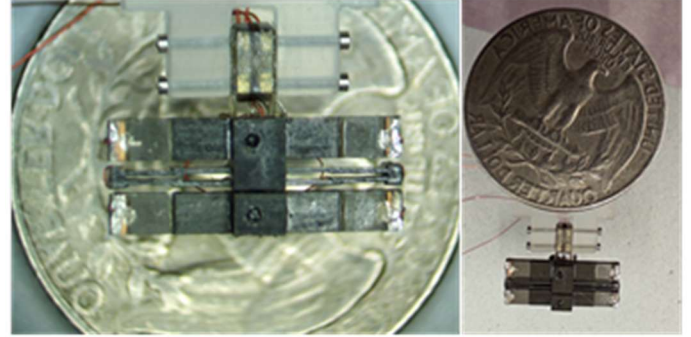


FIGURE 11: ASSEMBLED INCHWORM INSERTER COMPARED TO A U.S. QUARTER DOLLAR COIN.

The piezoelectric actuators' gripping force capability was also evaluated via the blocked clamping force. Free deflection (D_f) was measured with a microscope camera and motion tracking as a function of applied voltage. Blocked force ($F_{b,meas}$) was measured as a function of voltage with the actuator sandwiched between a rigid plate and a force sensor, as shown in Fig. 12. However, during the force tests, there was still some minor deflection (D_b) of the actuator due to a small initial gap between the actuator and the force sensor, which reduced the measured force. The estimated 'true' blocked force (F_b) was thus estimated by the equation:

$$F_b = \frac{F_{b,meas} \times D_f}{D_f - D_b} \quad (2)$$

Out of the nine fabricated inchworm actuators, when 250 V was applied to them, they were shown to be able to deliver up to 950 mN of blocked force output, as shown in Fig. 12(b). The three used in the current inchworm prototype were the three lowest average force of the nine actuators, yielding 948, 887, and 813 mN peak blocked force at 250 V activation voltage. To ensure consistency, all the inchworm tests were kept at 200 V or less. Also, the actual gripping force of the electrode would be smaller as the gap in between the grippers decreased. For example, if the gripper moved half of the free deflection before touching the electrode, it would provide half the blocked force. Thus, during actual inchworm tests, based on the 12 – 25 μm electrode thickness and the 200 V input, we estimated the upper and bottom gripper force to be around 755 mN and the vertical extender feeding force to be around 670 mN. Such force levels, while lower than the 950 mN ideal blocked force, were still sufficient for gripping the microelectrode to generate frictional forces for holding against the membrane dimpling and during the incremental insertion step.

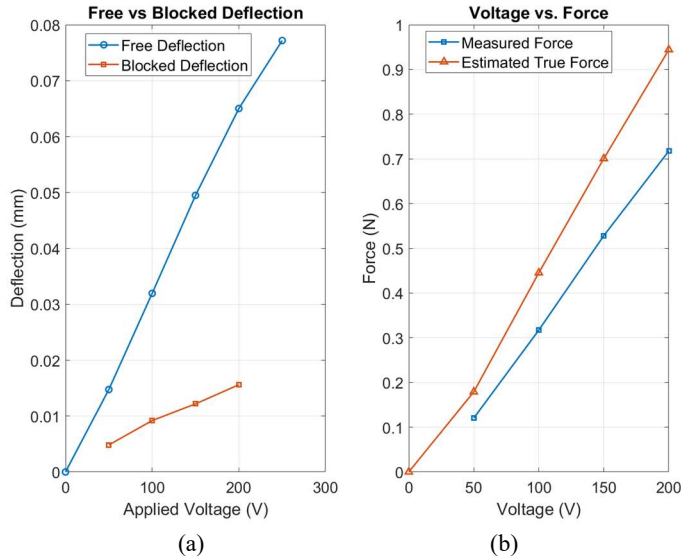


FIGURE 12: (A) FREE AND BLOCKED ACTUATOR DEFLECTION AS A FUNCTION OF APPLIED VOLTAGE WITH (B) COMPARISON OF MEASURED AND ESTIMATED BLOCKED FORCE RESPONSE.

The whole inchworm assembly bandwidth was characterized as up to 400 Hz, combined with the incremental insertion depth, yielding up to 28 mm/s linear insertion speed. Such experimental characterization results showed the capability of the inchworm insertion robot prototype, which was above the in vivo insertion measurement results for dura and pia mater penetrations (10s of mN level load) [33].

3.2 Inchworm Insertion Test Results

Insertion force profile processing results for insertion tests of 25 μm diameter bare tungsten microwires under various inchworm frequencies are as summarized in Fig. 13. With 67 μm per cycle insertion distance, the twelve inchworm insertion frequency levels were converted into average linear insertion speed ranging from 0.034 to 13.4 mm/s, assuming no slipping occurs during insertion and stable vertical extender motion during the insertion step against membrane resistance.

As can be seen in Fig. 13, the membrane rupture force was high at low inchworm frequencies. With the lowest 0.5 Hz frequency (0.034 mm/s linear speed) trial, the dura-pia rupture force was recorded at 10.96 mN, which was comparable to in vivo rat brain dura-pia measurement at 10.64 mN [33]. Similar match was observed for dimpling depth at rupture (Fig. 13(b)), yielding 0.917 mm dimpling at rupture for 0.5 Hz phantom insertion test, as compared to 0.919 mm dimpling from in vivo rat brain test under static stereotaxic insertions [33].

As the inchworm operation frequency increased, both the membrane rupture force and dimpling depth at rupture started to decline. The membrane rupture force minimized to 3.27 mN under 5.025 mm/s linear insertion speed, corresponding to 75 Hz inchworm operation frequency. Similarly, the membrane dimpling at rupture also reached the lowest level around 25 – 100

Hz (1.675mm/s - 6.7mm/s) range. While the specific mechanism of such reduction remained to be investigated, it was possible that the pecking motion enabled by the incremental inchworm insertion motion at higher linear speed provided extra kinematic energy to vary the brain membrane rupture mechanism, leading to a lower rupture and dimpling level at electrode penetration.

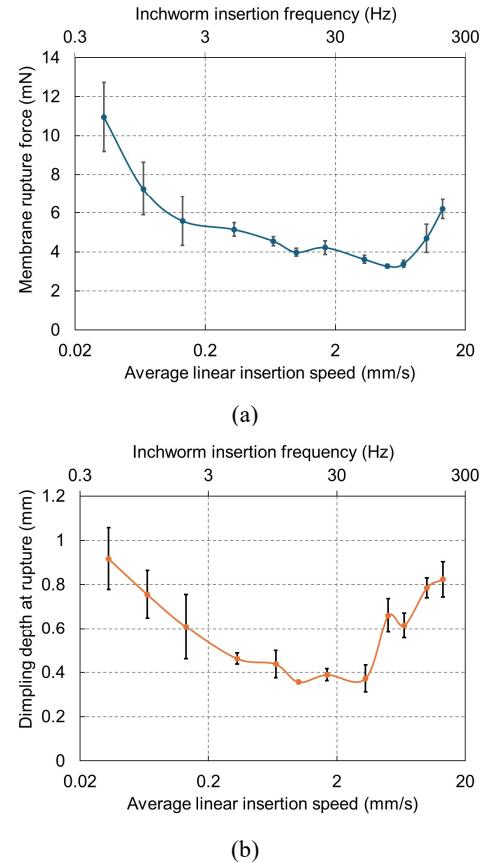


FIGURE 13: INSERTION TEST RESULTS OF 25 μm BARE TUNGSTEN MICROWIRE INSERTION INTO MULTI-LAYER BRAIN PHANTOM UNDER VARIOUS INCHWORM INSERTION FREQUENCIES. (A) THE MEMBRANE RUPTURE FORCE AND (B) MEMBRANE DIMPLING AT RUPTURE FORCE WERE PLOTTED AGAINST THE AVERAGE LINEAR INSERTION SPEED AND INCHWORM FREQUENCY IN LOG SCALE. ERROR BAR REPRESENTS STANDARD DEVIATION BETWEEN AT LEAST THREE REPEATED TRIALS UNDER EACH FREQUENCY.

However, at even higher frequencies beyond 75 Hz, it was observed that the membrane rupture force and dimpling depth started to rise again. Under high frequency (thus high linear insertion speed) insertion tests, the quick interaction between the grippers and the tungsten microwire might lead to unstable electrode holding of the electrode, causing slipping and unstable electrode-brain membrane interaction, which might be the reason for higher rupture force and dimpling. Such slippage could potentially be mitigated by (1) physically altering the design and decreasing the current 40 μm gap to match the clamped electrode, (2) swapping the gripper material for

improved frictional properties for better gripping, or (3) applying a closed loop feedback system that adjusts voltage signal real-time to provide more reliable gripping.

Furthermore, it could be noted that when operating around the optimal 75 Hz frequency (5.025 mm/s linear speed), deviations between tests were smaller than the lower or higher frequency counterparts, further validating the optimal and stable membrane cutting performance at that frequency and linear insertion speed. The 75 Hz optimal operation frequency also aligned with our previous experimental investigation of a mechanical cam-wheel driven inchworm prototype [17], which found that a moderate inchworm insertion speed might benefit the insertion force reduction and buckling prevention the most.

Based on the found optimal 75 Hz inchworm operating frequency, further insertion tests were conducted under this frequency for 12 μm diameter tungsten microwires and three different configurations of silicon probes as shown in Fig. 8. When inserted against the multi-layer brain phantom, all four devices penetrated the dura and pia layers without buckling, with Fig. 14 showing screenshots of their un-buckled status right before membrane rupture. Such results were convincing because these four types of electrodes of 15 μm thickness or less, were conventionally considered only viable for pia mater penetration, requiring a durotomy to remove partial dura mater tissue during surgery. This durotomy lead to more complicated procedures as well as higher risk of infection and further complications that affect chronic recording qualities. With the enhanced buckling-free insertion capability enabled by the developed inchworm insertion robot and the grip-insert-release incremental insertion motion, it was shown that all of them could be used towards dura penetration, potentially enabling high-density brain-wide implantation of such less-damaging neural interfaces for novel previously impossible neuroscience experiments.

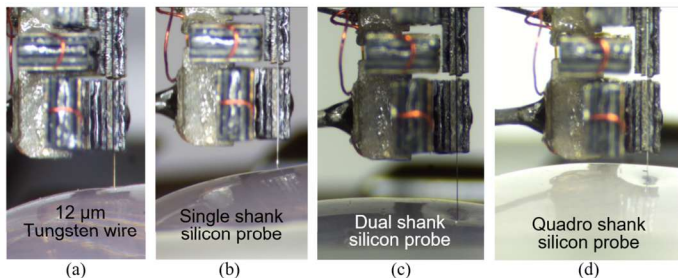


FIGURE 14: MEMBRANE RUPTURE POINT DURING SUCCESSFUL INSERTIONS INTO DURA-PIA LAYER PHANTOMS.

It is worth noting that while the proof-of-concept insertion trials showed promising results, the inchworm insertion mechanism needs further investigation in a few directions. Firstly, the exact cutting mechanism for the inchworm incremental insertion to minimize dimpling and membrane rupture force remains to be studied. Secondly, while 75 Hz, was determined as the optimal for the 25 μm , different optimal frequencies would likely be applied to other devices. For example, in our preliminary tests, single shank silicon probe

yielded smaller rupture force and dimpling depth under 50 Hz (3.35 mm/s), inchworm motion as compared to 75 Hz (5.025 mm/s). The inchworm motion frequency's impact on the insertion outcome as well as the overall inchworm operation control strategy for different neural interfaces need to be further investigated. Lastly, further optimization of the inchworm insertion robot design is needed to provide support for upper electronics of a full neural interface package, to further reduce the maximum unsupported length during inchworm operation, as well as to evaluate the inchworm inserter's impact on the chronic recording performance.

4. CONCLUSION

In this study, a piezoelectric actuator-based inchworm incremental insertion robot was designed and fabricated. Insertion tests against multi-layer brain mimicking phantom showed promising results enabling buckling-free penetration of dura and pia maters with 12 μm and 25 μm tungsten microwires as well as 15 μm thick silicon probes in single-, double-, and quad-shank configurations. The cyclic and incremental grip-insert-release inchworm motion had shown to reduce the membrane rupture force and dimpling depth at rupture as compared to conventional static stereotaxic insertions and minimized the electrode unsupported length to strengthen the electrode's critical buckling load. Thanks to the tunable properties of the piezo unimorph recurve actuators, the inchworm insertion concept for buckling prevention would be widely applicable to tissue insertion scenarios beyond the brain membrane penetration case. Future directions of the study include investigation of optimal process parameters for all types of electrodes, insertion dynamics and membrane rupture mechanics, gripper material and structural designs for various electrode surface coatings, inchworm insertion targeting accuracy, as well as in vivo verification of the inchworm inserter.

ACKNOWLEDGEMENTS

This study was financially supported by the National Science Foundation (NSF) under award number 2347299. The authors would like to thank the surgical input, feedback, and guidance provided by Drs. Brendon Watson and Jeremiah Hartner from the University of Michigan during the design, prototype, and testing of the inchworm insertion robot.

REFERENCES

- [1] Connors, B. W., and Regehr, W. G., 1996, "Neuronal firing: does function follow form?," *Curr Biol*, 6(12), pp. 1560-1562.
- [2] Shobe, J. L., Claar, L. D., Parhami, S., Bakhrin, K. I., and Masmanidis, S. C., 2015, "Brain activity mapping at multiple scales with silicon microprobes containing 1,024 electrodes," *J Neurophysiol*, 114(3), pp. 2043-2052.
- [3] Taylor, Z., and Miller, K., 2004, "Reassessment of brain elasticity for analysis of biomechanisms of hydrocephalus," *Journal of Biomechanics*, 37(8), pp. 1263-1269.
- [4] Yi, D., Yao, Y., Wang, Y., and Chen, L., 2024, "Design, Fabrication, and Implantation of Invasive Microelectrode Arrays

as in vivo Brain Machine Interfaces: A Comprehensive Review," *J Manuf Process*, 126, pp. 185-207.

[5] Muthuswamy, J., Saha, R., and Gilletti, A., "Tissue micromotion induced stress around brain implants," *Proc. 2005 3rd IEEE/EMBS Special Topic Conference on Microtechnology in Medicine and Biology*, pp. 102-103.

[6] Lecomte, A., Descamps, E., and Bergaud, C., 2018, "A review on mechanical considerations for chronically-implanted neural probes," *Journal of Neural Engineering*, 15(3), p. 031001.

[7] Someya, T., Bao, Z., and Malliaras, G. G., 2016, "The rise of plastic bioelectronics," *Nature*, 540(7633), pp. 379-385.

[8] Feron, K., Lim, R., Sherwood, C., Keynes, A., Brichta, A., and Dastoor, P. C., 2018, "Organic Bioelectronics: Materials and Biocompatibility," *International Journal of Molecular Sciences*, 19(8), p. 2382.

[9] Rajzer, I., Menaszek, E., Bacakova, L., Rom, M., and Blazewicz, M., 2010, "In vitro and in vivo studies on biocompatibility of carbon fibres," *Journal of Materials Science: Materials in Medicine*, 21(9), pp. 2611-2622.

[10] Petersen, R., 2016, "Carbon Fiber Biocompatibility for Implants," *Fibers*, 4(1), p. 1.

[11] Briggs, A., 1977, "Carbon fibre-reinforced cement," *Journal of Materials Science*, 12(2), pp. 384-404.

[12] Zhao, Q., Zhang, K., Zhu, S., Xu, H., Cao, D., Zhao, L., Zhang, R., and Yin, W., 2019, "Review on the Electrical Resistance/Conductivity of Carbon Fiber Reinforced Polymer," *Applied Sciences*.

[13] Rodger, D. C., Fong, A. J., Li, W., Ameri, H., Ahuja, A. K., Gutierrez, C., Lavrov, I., Zhong, H., Menon, P. R., Meng, E., Burdick, J. W., Roy, R. R., Edgerton, V. R., Weiland, J. D., Humayun, M. S., and Tai, Y.-C., 2008, "Flexible parylene-based multielectrode array technology for high-density neural stimulation and recording," *Sensors and Actuators B: Chemical*, 132(2), pp. 449-460.

[14] Guo, L., Guvanasen, G. S., Liu, X., Tuthill, C., Nichols, T. R., and DeWeerth, S. P., 2013, "A PDMS-Based Integrated Stretchable Microelectrode Array (isMEA) for Neural and Muscular Surface Interfacing," *IEEE Transactions on Biomedical Circuits and Systems*, 7(1), pp. 1-10.

[15] Marelli, M., Divitini, G., Collini, C., Ravagnan, L., Corbelli, G., Ghisleri, C., Gianfelice, A., Lenardi, C., Milani, P., and Lorenzelli, L., 2011, "Flexible and biocompatible microelectrode arrays fabricated by supersonic cluster beam deposition on SU-8," *Journal of Micromechanics and Microengineering*, 21(4), p. 045013.

[16] Cheng, M.-Y., Park, W.-T., Yu, A., Xue, R.-F., Tan, K. L., Yu, D., Lee, S.-H., Gan, C. L., and Je, M., 2013, "A flexible polyimide cable for implantable neural probe arrays," *Microsystem Technologies*, 19(8), pp. 1111-1118.

[17] Yi, D., Landry, N., Blake, S., Baron, J., and Chen, L., 2024, "An Experimental Study of Incremental Buckling-Resistant Inchworm-Type Insertion of Microwire Neural Electrodes," *Journal of Manufacturing Science and Engineering*, 146(11).

[18] Chen, S., and Kucernak, A., 2002, "Fabrication of carbon microelectrodes with an effective radius of 1 nm," *Electrochemistry Communications*, 4(1), pp. 80-85.

[19] Millar, J., and Pelling, C. W., 2001, "Improved methods for construction of carbon fibre electrodes for extracellular spike recording," *J Neurosci Methods*, 110(1-2), pp. 1-8.

[20] Salcman, M., and Bak, M. J., 1973, "Design, fabrication, and in vivo behavior of chronic recording intracortical microelectrodes," *IEEE Trans Biomed Eng*, 20(4), pp. 253-260.

[21] Guitchounts, G., Markowitz, J. E., Liberti, W. A., and Gardner, T. J., 2013, "A carbon-fiber electrode array for long-term neural recording," *Journal of Neural Engineering*, 10, p. 046016.

[22] Dong, T., Chen, L., and Shih, A., 2020, "Laser Sharpening of Carbon Fiber Microelectrode Arrays for Brain Recording," *J Micro Nanomanuf*, 8(4), p. 041013.

[23] Na, K., Sperry, Z. J., Lu, J., Voroslakos, M., Parizi, S. S., Bruns, T. M., Yoon, E., and Seymour, J. P., 2020, "Novel diamond shuttle to deliver flexible neural probe with reduced tissue compression," *Microsyst Nanoeng*, 6, p. 37.

[24] Kim, H.-J., Clement, R. S., Bagwell, R. B., Natasha, T. N., Shih, Y.-Y. I., and Lee, S.-H., "Direct Mapping of the Nucleus Accumbens Core and Shell using Deep Brain Stimulation with functional Magnetic Resonance Imaging in Rats," *Proc. ISMRM & SMRT Annual Meeting & Exhibition*.

[25] Sharp, A. A., Ortega, A. M., Restrepo, D., Curran-Everett, D., and Gall, K., 2009, "In Vivo Penetration Mechanics and Mechanical Properties of Mouse Brain Tissue at Micrometer Scales," *IEEE Transactions on Biomedical Engineering*, 56, pp. 45-53.

[26] Patel, P. R., Na, K., Zhang, H., Kozai, T. D. Y., Kotov, N. A., Yoon, E., and Chestek, C. A., 2015, "Insertion of linear 8.4 μm diameter 16 channel carbon fiber electrode arrays for single unit recordings," *Journal of Neural Engineering*, 12, p. 046009.

[27] Hanson, T. L., Diaz-Botia, C. A., Kharazia, V., Maharbiz, M. M., and Sabes, P. N., 2019, "The "sewing machine" for minimally invasive neural recording," *bioRxiv*, p. 578542.

[28] Musk, E., and Neuralink, 2019, "An Integrated Brain-Machine Interface Platform With Thousands of Channels," *J Med Internet Res*, 21(10), p. e16194.

[29] Khilwani, R., Gilgunn, P. J., Kozai, T. D., Ong, X. C., Korkmaz, E., Gunalan, P. K., Cui, X. T., Fedder, G. K., and Ozdoganlar, O. B., 2016, "Ultra-miniature ultra-compliant neural probes with dissolvable delivery needles: design, fabrication and characterization," *Biomed Microdevices*, 18(6), p. 97.

[30] Chen, L., Hartner, J., Van Dyke, D., Dong, T., Watson, B., and Shih, A., "Custom skull cap with precision guides for deep insertion of cellular-scale microwire into rat brain," *Proc. International Manufacturing Science and Engineering Conference, American Society of Mechanical Engineers*, p. V001T005A009.

[31] Smith, R. D., Kolb, I., Tanaka, S., Lee, A. K., Harris, T. D., and Barbic, M., 2022, "Robotic multi-probe single-actuator inchworm neural microdrive," *eLife*, 11, p. e71876.

[32] Abondance, T., Jayaram, K., Jafferis, N. T., Shum, J., and Wood, R. J., 2020, "Piezoelectric Grippers for Mobile Micromanipulation," *IEEE Robotics and Automation Letters*, 5(3), pp. 4407-4414.

[33] Chen, L., Hartner, J., Dong, T., Li, A., Watson, B., and Shih, A., 2021, "Flexible High-Resolution Force and Dimpling Measurement System for Pia and Dura Penetration During In Vivo Microelectrode Insertion Into Rat Brain," *IEEE Trans Biomed Eng*, 68(8), pp. 2602-2612.

[34] Jafferis, N. T., Jayaram, K., York, P. A., and Wood, R. J., 2021, "A streamlined fabrication process for high energy density

piezoelectric bending actuators," *Sensors and Actuators A: Physical*, 332.

[35] Yi, D., Lat, K., and Chen, L., 2026, "Multi-Layer Brain-Mimicking Phantom for Replicating Dura and Pia Membrane Dimpling and Rupture Properties During Neural Interface Implantation," *bioRxiv*, p. 2026.2002.2005.704082.