Bending Properties of Materials for Peripheral Nerve Interfaces

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Abstract—Intrafascicular peripheral nerve interfaces (PNIs) with penetrating electrodes have the potential to chronically record from nerves on the scale of single axons. The small size and dynamic environment of peripheral nerves makes material selection important for these devices. In this work, we describe how the bending properties of common PNI electrode materials contribute to their effectiveness as self-inserting PNIs. First, tungsten, platinum-iridium, and carbon fiber wires are tested to assess their ability to survive repeated bending stresses when embedded in silicone. Next, carbon fiber wires are attached to a flexible circuit board encased in silicone to characterize how they survive stresses in prototype PNI devices. Finally, in order to validate experimental results, we use COMSOL to investigate the optimal thickness of the embedded silicone layer by simulating the stress distribution in carbon fiber wires on a flexible circuit board. Carbon fiber wires were shown to survive bending stresses better than tungsten and platinum-iridium wires. Physical testing and COMSOL modeling of carbon fiber prototype devices showed an optimal silicone thickness of 200 μm that prevents carbon fiber breakage but minimizes PNI device size. Overall, these results serve as a guide for selection of self-inserting PNI materials and development of carbon fiber PNIs.

I. INTRODUCTION
Peripheral nerve interfaces (PNIs) allow us to access the nervous system in order to understand and solve issues that can arise because of disease or injury. PNIs have been implemented in many forms, but consist of two main types: intrafascicular and extrafascicular. Intrafascicular interface devices place contacts outside of the epineurium and are minimally invasive but do not have the ability to record from individual axons. Examples of extrafascicular devices include the cuff electrode [1] [2], flat-interface nerve electrode (FINE) [3], and book electrode [4]. Intrafascicular interface devices penetrate the epineurium and place electrode contacts inside the fascicles. These devices have the potential to chronically record from nerves at a single axon level. Examples of previously developed intrafascicular interface devices include the longitudinal intrafascicular electrode (LIFE) [5], transverse intrafascicular multichannel electrode (TIME) [6], [7], Utah array [8], and floating microelectrode array (FMA) [9], [10].

Flexible polymer based devices like the LIFE and TIME are made out of the LIFE and TIME are inserted into the fascicle using a stiff external needle [5], [6]. The needle is then removed, leaving the polymer based device behind in the fascicle. This approach allows soft materials to be left in contact with the nerve, but the process of implantation is difficult. Penetrating microelec trodes like the Utah array and FMA are arrays of small diameter wire or needle shaped structures that are stiff enough to insert into the nerve. These devices are created with a limited number of materials because of the unique difficulties of their insertion process and chronic environment. Penetrating microelectrodes must be small enough to penetrate a peripheral nerve while causing minimal damage, implant in areas that are not easily accessible, and last for long periods of time in areas that are constantly moving. The Utah array, made of silicon and developed for central nervous system applications, has been used in peripheral nerves [8], but has been shown to cause scarring [11], [12], and other silicon arrays have proven difficult if not impossible to insert into peripheral nerves [13]. For these reasons, novel materials may be better than silicon for peripheral nerve applications. Materials that have been used both in the brain and for intrafascicular device electrodes include tungsten [14], platinum-iridium [15], and carbon fiber [16], [17], [18].

This paper addresses the mechanical properties of tungsten (W), platinum-iridium (PtIr), and carbon fiber (CF) wires in order to determine which are most suitable for use as penetrating microelectrodes for PNIs. We characterized the bending of carbon fiber wires in prototype devices both experimentally and with simulation. We found that carbon fibers can survive cyclical stress better than other wires, which is desirable for implantation and recording in the demanding environment of peripheral nerves. We also found that embedding carbon fibers in silicone improves their ability to survive stresses associated with implantation in peripheral nerves and there appears to be an optimal thickness of silicone that minimizes breakage while still minimizing overall device size.

II. METHODS

In this work, we first created bend test devices with wires made out of W, PtIr, and CF embedded in a silicone block.
We then characterized the failure of each type of wire in these devices with a bend test. Next, we characterized the bending of CF wires in prototype devices, when attached to a flexible circuit board and embedded in silicone. Finally, we used COMSOL Multiphysics to simulate the stress in CF wires when attached to a flexible circuit board and embedded in silicone in order to validate experimental results.

A. Preparation of Bend Test Devices

Bend test devices were designed to test the bending characteristics of W (100211, California Fine Wire, Grover Beach, CA), PtIr (100167, California Fine Wire, Grover Beach, CA), and CF (T-650/35 3K, Cytec Thornel, Woodland Park, NJ) wires when embedded in medical-grade silicone (A-103, Factor II, Lakeside, AZ). We designed and created a custom setup that consists of 3 pieces: a baseplate, an alignment board, and an alignment board holder. Using these 3 pieces, we were able to consistently embed individuated wires spaced evenly and centered in uniform pieces of silicone.

The baseplate was used to hold silicone in the desired shape while it cured. We designed and ordered a CNC milled aluminum baseplate (Protolabs, Maple Plain, MN) with 6 1.8 mm long by 1 mm wide by 0.6 mm deep "wells" in the surface. The baseplate has 4 round holes surrounding each well used to keep the alignment board holder in place. Custom printed circuit board (PCB) alignment boards were used to hold 8 wires at a pitch of 152.4 μm in place within the desired baseplate well while the silicone cured. Wires were secured to alignment boards and cut so they extended 1.75 mm past the edge of the board. Holders for the alignment boards were designed in CAD software and printed with a Form 2 SLA 3D printer (Formlabs, Somerville, MA). The holders were designed with pins that correspond with the round holes in the baseplate to precisely hold the alignment boards such that the wires are centered along both axes of the silicone well.

Figure 1 shows the setup used to create these devices. First, alignment boards with wires were secured to the alignment board holder with carbon tape (16073, Ted Pella, Redding, CA). Next, the baseplate wells were filled with silicone so that they were overflowing with silicone then degassed in a vacuum chamber for 20 minutes. The silicone was then cured at 110°C for 20 minutes. Figure 2 shows one of these devices set up for a bend test.

B. Bend Testing of Selected Materials

In order to simulate bending stresses experienced by PNIs, we developed a method of repeatedly bending the embedded wires. In brief, bend test devices were cut to length, placed on a flat level surface, and wires were bent with a glass capillary mounted on a linear actuator.

Bend test devices were prepared as described above with CF, W, and PtIr wires. Wires were manually cut to ~150 μm above the silicone for bend test devices and ~250 μm for prototype devices. These devices were placed on a level surface and secured with double sided carbon tape. A precision linear actuator (M-235.5DD, PI, Auburn, MA) was mounted to a micromanipulator and aligned such that the actuation was perpendicular to the wires and parallel to the surface of the silicone. A 0.75 mm diameter glass capillary (625000, A-M Systems, Sequim, WA) was secured to the end of the linear actuator with a custom 3D-printed part.

A digital microscope was set such that the entire bend test device and capillary were in view, and a ruler set in the frame for scale. Using the microscope and ImageJ, the height of the glass capillary above the silicone was adjusted. For all tests the capillary was set between 40 and 60 μm above the surface

![Fig. 1. Setup for fabricating bend test devices. CF wires are held in a well filled with silicone while the silicone cures.](image)

![Fig. 2. An example of the setup for bend tests. (Left) A bend test device with CF wires. (Right) A prototype device with CF wires and the capillary that is used to bend wires. The polyimide "board" is outlined for clarity.](image)
of the silicone and passed over the wires 500 times. The wires were checked at 100 cycles and 500 cycles. In this paper, we define 1 cycle as the capillary sweeping the surface of all wires and returning to its starting position. Pictures of the wires were taken at time points of interest. Figure 2 shows an example of the bend test setup.

C. Modeling of Prototype Devices

In order to understand the impact of the silicone layer thickness on CF wire bending, we developed a model of these devices in COMSOL Multiphysics®. The model consists of a polyimide board with a CF wire fixed to it, encased in silicone. Following the experimental setup, the polyimide board is modeled by a 0.375 mm tall by 1.875 mm long by 0.100 mm thick rectangular prism. An 8.4 μm semi-cylindrical section of the board is cut out of the long side of the board from top to bottom as a slot for the CF to simulate bonding with an epoxy in the experimental setup. The CF is an 8.4 μm diameter cylinder, the diameter of a CF with an average coating of parylene-c, that is placed in the slot with one end flush to the bottom of the polyimide and the other end extending past the top edge of the board. The silicone is modeled as two variable height by 1.875 mm long by 0.200 mm thick rectangular prisms. The board and CF are sandwiched between these two pieces of silicone. The length of the CF and height of the silicone are varied together in all simulations such that the CF always extends 250 μm past the edge of the silicone. This creates a single fiber device with similar dimensions to those described in the prototype devices section. The material properties of each component are listed in Table I.

<table>
<thead>
<tr>
<th>Material</th>
<th>Property</th>
<th>Value</th>
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<tbody>
<tr>
<td>Silicone</td>
<td>Density [19]</td>
<td>1.12 g/cm³</td>
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<tr>
<td></td>
<td>Young Modulus [20]</td>
<td>1.01 MPa</td>
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<tr>
<td></td>
<td>Poisson’s Ratio [19]</td>
<td>0.49</td>
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<tr>
<td>Polyimide</td>
<td>Density [21]</td>
<td>1.42 g/cm³</td>
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<tr>
<td></td>
<td>Young Modulus [21]</td>
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<tr>
<td></td>
<td>Poisson’s Ratio [21]</td>
<td>0.34</td>
</tr>
<tr>
<td>Carbon Fiber</td>
<td>Density [22]</td>
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</tr>
<tr>
<td></td>
<td>Young Modulus [23]</td>
<td>241 GPa</td>
</tr>
<tr>
<td></td>
<td>Poisson’s Ratio [22]</td>
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A static stress analysis was conducted in COMSOL. Contact areas between the CF and the board are assumed to be rigidly connected by epoxy and the corresponding surfaces were tied together in the model. The silicone-to-board and silicone-to-CF interfaces are also assumed to be rigidly connected as no obvious relative displacement was observed in experiments along these surfaces. The bottom of the board was globally fixed in place to simulate the fixation of the bottom of the device to carbon tape. To model bending, a prescribed displacement of 100 μm is applied to the tip of the CF. The von Mises stress at the board-CF interface and at the point where the CF enters the silicone were extracted from the model. The entire setup is shown in Figure 3.

III. RESULTS

A. Bend Testing with Different Materials

We performed bend tests with bend test devices made using PtIr, W, and CF wires. Devices with different types of wires were all fabricated and tested with identical processes in order to ensure consistency. During testing, wires were bent to ~70° with a glass capillary as described in the methods.

Failure modes for each wire type appeared differently. Figure 4 shows representative failure for each wire type when embedded in silicone. PtIr and W wires showed plastic deformation, a change in shape under stress such that the original shape of the wires was not recovered after the stress was removed. CFs showed brittle fracture, fracture that is preceded by little or no plastic deformation.

![Fig. 3. COMSOL model of prototype devices. A polyimide board with an attached CF is embedded in a silicone block. The CF extends 250 μm past the edge of the silicone. Displacement of the fiber tip is 100 μm parallel to the long edge of the board.](image)

![Fig. 4. Comparison of how different materials partially embedded in silicone deform when bent to ~70° from vertical. (Top) Carbon fiber wires. (Middle) Tungsten wires. (Bottom) Platinum-iridium wires.](image)
Figure 5 shows the angle of deflection of wires in these bend tests at 100 and 500 cycles. None of the CF wires (n = 24) showed any deflection through all 500 cycles. At 100 cycles, W wires (n = 27) showed a median deflection angle of 24.56°. The median deflection of W wires increased to 28.23° at 500 cycles. At 100 cycles, PtIr wires (n = 23) showed a median deflection of 53.02°. PtIr wires did not show any change between 100 and 500 cycles, likely because the wires were already bent beneath the height of the glass capillary.

We performed an extensive fatigue test on a limited number of bend test devices with CF wires to determine if the CFs would break when subjected to this bending stress. One device with n = 8 CFs was run to 25000 passes, checking CFs at regular intervals. Between 1000 and 3000 passes one CF failed, as seen in Figure 4 (Top). The other 7 CFs did not fail through 25000 passes.

**B. Further Characterization of Carbon Fiber Bending**

Unlike W and PtIr, CF wires were able to bend 25000 times to ~70° from vertical with minimal failure when embedded in silicone. To make devices as small as possible, thereby reducing the weight and invasiveness of PNI devices made with CF wires and silicone, we performed bend testing with prototype devices to see how much silicone was needed above the CF-to-silicone interface to prevent failure of CF wires.

Figure 6 shows the results of the bend testing with prototype devices. Wires with a silicone height of 0-100 μm above the edge of the board showed 57.14% failure of CF wires (n = 4/7), 100-150 μm showed 25% failure (n = 3/12), 150-200 μm showed 12.5% failure (n = 1/8), 200-250 μm showed 10% failure (n = 1/10), and 250-300 μm showed 0% failure (n = 0/6). These results suggest that CFs are less likely to fail when there is more silicone above the connection point between the fibers and the board. CFs show a high percentage of failure when silicone is less than 150 μm above the edge of the board. Above this thickness, increasing silicone thickness comes with a minimal decrease in failure of CF wires.

**C. COMSOL Analysis of Carbon Fiber Bending**

To gain a better idea of the optimal thickness of silicone above the edge of the board, we developed a COMSOL model based on the prototype devices. Single CF wires were modeled as fixed to a polyimide board and embedded in silicone. The von Mises stress in the fiber was extracted at the connection point with the board and at the surface of the silicone. The goal of this model was to understand how stress in the CF changes at two points of interest: the connection point of the wire to the polyimide board and the surface of the silicone. The model and physical testing are visually similar (Figure 7), confirming that the model is useful for understanding the stresses in the physical testing.
The simulation result of CF bending inside a 200 μm thick silicone layer is as shown in Figure 7 (right). Two peak values of von Mises stress were found along the CF at the connection points of the wire to the polyimide board and the surface of the silicone. As stress concentration points, these two locations are also critical for fatigue life analysis and understanding the impact of silicone layer thickness on the CF bend test performance.

The model shows that as the silicone thickness above the edge of the board increases, the stress at the edge of the board and the edge of the silicone decreases (Figure 8). With 0 μm of silicone, the stress at the edge of the silicone and at the edge of the board are the same value because the edge of the silicone and edge of the board are the same point. Between 0 and 200 μm of silicone the stress at the edge of the board is higher than the stress at the surface of the silicone. At 200 μm of silicone, the stress at the edge of the board is slightly lower than the stress at the surface of the silicone. With more than 200 μm of silicone, the stress at the surface of the silicone begins to decrease at a lower rate, causing it to overcome the stress at the board. Increasing the thickness of the silicone beyond 200 μm shows a diminishing return. This is consistent with the physical testing of CFs (Figure 6).

IV. DISCUSSION

In this project, we analyzed the bending properties of materials embedded in silicone for use in PNiS. Our results suggest three main findings. First, CFs have many favorable properties for implantation and recording in the demanding environment of peripheral nerves. Peripheral nerves are generally not easily accessible, so devices implanted in peripheral nerves must be able to withstand large stresses due to inadvertent contact with other tissue during surgery. Many peripheral nerves are also constantly moving, so devices must be able to withstand repeated stresses once implanted. Our tests showed that CFs can undergo repeated cyclical stress without plastic deformation better than W and PtIr wires.

Next, embedding CFs in silicone improves their ability to survive stresses associated with implantation in peripheral nerves. While we have seen in other situations that CFs may buckle [24], our results suggest that they can bend without breaking for many cycles. We found experimentally and through modeling that adding silicone above the edge of the circuit board-to-CF interface decreases the stress on the CF and therefore decreases the likelihood of the CF breaking.

Finally, there appears to be an optimal thickness of silicone that minimizes breakage while still minimizing size. In order to use PNiS in upper extremities of humans, they should be able to be implanted and connected to nerves that are as small as 1 mm in diameter [25]. Both our experimental results and models show that about 100-200 μm of silicone above the edge of the circuit board-to-CF interface decreases breakage while minimizing size. This would allow PNiS made with silicone and CF wires to be attached to ~1 mm peripheral nerves without being much larger than the nerve.

Future work could be done to further understand how to best use CF wires for PNiSs. Testing to further characterize breakage and stress in physical situations could be done with the free end of CF wires from prototype devices embedded in a moving nerve. Different modes of microscopic fracture and failure in CF wires could be analyzed to understand and minimize breakage. Material properties of different CF wires could be compared to choose the best CF wires for use in PNiSs.

REFERENCES


