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Mechanical failure modes of chronically implanted planar silicon-based neural probes for laminar recording

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ABSTRACT

Penetrating intracortical electrode arrays that record brain activity longitudinally are powerful tools for basic neuroscience research and emerging clinical applications. However, regardless of the technology used, signals recorded by these electrodes degrade over time. The failure mechanisms of these electrodes are understood to be a complex combination of the biological reactive tissue response and material failure of the device over time. While mechanical mismatch between the brain tissue and implanted neural electrodes have been studied as a source of chronic inflammation and performance degradation, the electrode failure caused by mechanical mismatch between different material properties and different structural components within a device have remained poorly characterized. Using Finite Element Model (FEM) we simulate the mechanical strain on a planar silicon electrode. The results presented here demonstrate that mechanical mismatch between iridium and silicon leads to concentrated strain along the border of the two materials. This strain is further focused on small protrusions such as the electrical traces in planar silicon electrodes. These findings are confirmed with chronic *in vivo* data (133–189 days) in mice by correlating a combination of single-unit electrophysiology, evoked multi-unit recordings, electrochemical impedance spectroscopy, and scanning electron microscopy from traces and electrode sites with our modeling data. Several modes of mechanical failure of chronically implanted planar silicon electrodes are found that result in degradation and/or loss of recording. These findings highlight the importance of strains and material properties of various subcomponents within an electrode array.

1. Introduction

Chronically implanted intracortical electrode arrays are powerful research tools for understanding functions of the brain [1–11]. These tools are used for monitoring neural activity longitudinally to study memory, plasticity, and aging, and can be especially powerful when combined with emerging transgenic and *in vivo* imaging tools [12]. They have also demonstrated promising clinical use in allowing human patients with tetraplegia to control neuroprosthetic devices such as a robotic arm or computer cursors [13,14]. While long-term neural implants have demonstrated feasibility [13–16], the large variability and poor longevity of the recorded signals have presented a major challenge [17–23]. Electrophysiology and histology results show significant differences in recording performance and tissue response with the same device across different animals, between different electrode shanks in the same animal, and at different depths on the same shank [17–19,24–28]. This performance variability and degradation are understood to be the result of a complex combination of biological and material failure mechanisms [20,22].

Many recent studies have focused on understanding the biological sources of variability. During implant insertion, penetrating a single major intracortical blood vessel results in a significantly large area of bleeding and blood brain barrier (BBB) disruption

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when compared to penetrating through many small capillaries [17]. This bleeding damages the local tissue, diminishing recording performance [29]. It has also been shown that implanting electrodes closer to major penetrating blood vessels, without damaging them, leads to increased astroglial activity [30,31]. Consequences of BBB damage are not limited to leakage, the loss of oxygen perfusion to the tissue near the probe can also lead to secondary metabolic injury [12]. In addition to the initial insertion injury [17,32,33], chronic presence of the implant may cause persistent BBB disruption, chronic inflammation, and neuronal degeneration, which may lead to long-term recording failure of the implanted electrodes [20,29].

While biological tissue responses contribute to the recording variability and degradation, electrode material failure also plays a significant role. Currently, the most widely used and commercially available intracortical electrode arrays for chronic neural recordings include bed-of-needle type (eg. microwire arrays and bulk silicon micromachined Utah-arrays) [18,34], and thin-film microfabricated planar arrays [35]. Several studies have characterized the material failure mechanism of microwire- and Utah-arrays [20,36,37]. Failure modes include corrosion, cracking, bending of recording sites, and delamination or cracking of the insulating polymer materials, such as parylene-C [20,36,37]. However, limited investigation has been reported on the mechanical failure modes of thin-film planar silicon arrays. These arrays can have multiple electrode sites located along the depth of the probe shanks, which allows for the simultaneous sampling of multiple cortical and subcortical layers necessary for many neural circuitry studies [1,38–40]. Understanding the failure modes of these arrays will help identify limitations in research and applications as well as potential opportunities for design improvements.

One representative planar array that is commercially available and widely used is the silicon based Michigan probes. These electrodes are made on a degenerately Boron-doped silicon-on-insulator substrates (Fracture Strength: 1800 MPa) [35,41,42]. An insulating layer of silicon oxide is deposited onto the substrate (Fracture Strength: 360 MPa) [43], which is followed by the deposition of conductive polycrystalline silicon (polysilicon) electrode traces (Fracture Strength: 1200 MPa) [44]. The traces are then insulated with a layer of silicon oxide, or a multilayer silicon oxide; silicon nitride; silicon oxide film. Finally, the iridium recording site is sputter coated on the electrode contact pads (see Refs. [35,41] for details) (Fracture Strength: 500–740 MPa) [45].

While chronically implanted planar silicon arrays are widely used in chronic rodent neuroscience studies [46–48], challenges have been reported in chronic primate applications [49,50]. These issues are likely due to the increase in brain size, increased interstitial space in the skull for brain movement, and the consequent increase in force experienced by the probes. Finite Element Modeling (FEM) studies have explored the tissue strain resulting from the mechanical mismatch between the planar probe and tissue [51,52]. It has also been shown that this mechanical mismatch in the brain changes over time [53]. However, the strain resulting from mechanical mismatch within a device has largely been ignored.

This study aims to understand the mechanical failure modes of silicon based planar electrode arrays. Here we employ FEM to examine the microstructures of the array that are vulnerable to mechanical strain. We show that the protruded electrode traces are focal points of induced mechanical strain, particularly around the edges of the iridium recording sites where mechanical strain is highest. The results are then validated by electrodes that were implanted in the visual cortex of mice for 4–6 months. Electrode performance degradation is assessed through impedance spectroscopy and neural recordings, which are correlated to visible polysilicon trace damage near the iridium recording site observed by scanning electron microscopy (SEM). Lastly, the interstitial space between the brain and skull are examined in human, primate, and rodent via MRI to estimate the extent of the mechanical failure with the scaling of animal size and provide insight towards future microelectrode design.

2. Methods

2.1. Finite-element model

A 3D finite-element model of a 15 μm thick, 123 μm wide planar silicon electrode was developed in ANSYS 14.5 (Canonsburg, PA) to examine von Mises Equivalent Elastic Strain using a model previously established [52]. Briefly, von Mises Equivalent Elastic Strain represents the effective strain on an object combined from three principal strains in mutually perpendicular axes and best characterizes the total strain along the entire probe. This is represented in ANSYS by the formula;

$$\varepsilon_e = 1\frac{1}{1+\nu'}\sqrt{\frac{1}{2}\left[(\varepsilon_1-\varepsilon_2)^2+(\varepsilon_2-\varepsilon_3)^2+(\varepsilon_3-\varepsilon_1)^2\right]} \quad [1]$$

where ε_e represents the von Mises Equivalent Elastic Strain, ν' is the effective Poisson's ratio, and ε_1, ε_2, and ε_3 represents the principal strains oriented to the three axes [54].

16 electrical traces were extruded 0.6 μm from the surface of the electrode with 2.5 μm trace width and 4 μm inter-trace spacing using the 'add material' feature. To simplify the model, the polysilicon traces, silicon oxide or silicon oxide/silicon nitride insulation were modeled with the same material properties as bulk silicon (Young's modulus of 200 GPa and Poisson's ratio of 0.278) used in a previously published model [52]. It is estimated that this model will be a conservative evaluation as polysilicon, silicon oxide, and silicon nitride are considered to be more brittle than the bulk silicon.

To model 703 μm^2 single shank Michigan electrodes, 16 iridium (Ir) recording sites with 30 μm diameter were protruded 1500 Å from the traces using the 'add frozen' operation. Ir material properties were imported from the default ANSYS Engineering Data Library (Young's modulus of 528 GPa and Poisson's ratio of 0.26) [55]. The recording sites and substrate/trace parts were combined into one body. This simplifies the model by assuming that the recording sites will not delaminate from the substrate.

Again, using the previously published model, a 1.4 mm × 1.4 mm × 2.0 mm cube at 37 °C was used to model the brain (Young's modulus of 6 kPa and Poisson's ratio of 0.45) [52]. The probe was modeled as initially starting in the center of the brain, implanted to a depth of 1.65 mm. In addition, mechanical strain was applied similarly by fixing the bottom surface of the brain cube and applying a 1 μm displacement in the thickness and width directions to model micromotion [52]. This simulation was used to determine the impact of the implanted electrode on tissue strain (Fig. 1a,b).

Due to large differences in modulus between the brain tissue and the probe, ANSYS could not accurately determine the strain on the probe when simultaneously calculating the strain on the brain tissue. Therefore, the mechanical strain on the probe was modeled without the brain phantom; instead the strain was applied directly to the tip of the electrode (Fig. 1c–e). Previous models, as well as the above brain-electrode model showed that the greatest strain in the tissue occurs at the tip of the probe (Fig. 1b). This was modeled by fixing the electrode base and applying the same tip displacement as in the previous model, assuming forces between probe and tissue balance (Newton's third law).

2.2. Surgical implantation

Single shank Michigan electrodes were implanted unilaterally into the left primary monocular visual cortex (V1m), of 9 wk old C57B6-Casp1 Knockout ($n = 3$) and age matched C57B6 wildtype ($n = 3$) female mice (22–28 g) [56]. The knockout of animals were used for studying the molecular and cellular pathways that mediate neural tissue response to implants and the results are reported in our companion study [28]. While half of the examined electrodes were explanted from KO mice, implants in KO and WT animals are considered equivalent and mechanical failure modes of the electrode is not expected to be different across animal types. Each animal was anesthetized under 1.5% isoflurane and mounted onto a stereotaxic frame (Kopf Instruments, Tujunga, CA). The top surface of the skull was exposed and a drill sized craniotomy was made centered at 1 mm anterior to Lambda and 1.5 mm lateral to midline using a high-speed dental drill and a 0.007 drill bit. Saline was applied continuously onto the skull to dissipate heat from the high-speed drill. Extra care was taken to prevent damage to the dura by reducing the drill speed and gently manually feeling the resistance of the skull when the dural blood vessels become visible through the opaque thin skull. A total of three bone screws were installed bilaterally over the primary motor cortex as well as over the contralateral visual cortex. The reference wire was connected to the bone screw over the contralateral visual cortex, while the ground wire was connected to both bone screws over the motor cortex. Arrays were inserted at ~2 mm/s using a stereotaxic manipulator until