

Magnetic insertion system for flexible electrode implantation

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ABSTRACT

Chronic recording electrodes are a vital tool for brain research and neural prostheses. Despite decades of advances in recording technology, probe structures and implantation methods have changed little over time. Then as now, compressive insertion methods require probes to be constructed from hard, stiff materials, such as silicon, and contain a large diameter shank to penetrate the brain, particularly for deeper structures. The chronic presence of these probes results in an electrically isolating glial scar, degrading signal quality over time. This work demonstrates a new magnetic tension-based insertion mechanism that allows for the use of soft, flexible, and thinner probe materials, overcoming the materials limitations of modern electrodes. Probes are constructed from a sharp magnetic tip attached to a flexible tether. A pulsed magnetic field is generated in a coil surrounding a glass pipette containing the electrode. The applied field pulls the electrode tip forward, accelerating the probe into the neural tissue with a penetration depth that is calibrated against the charge voltage. Mathematical modeling and agar gel insertion testing demonstrate that the electrode can be implanted to a predictable depth given system specific parameters. Trial rodent implantations resulted in discernible single-unit activity on one of the probes. The current prototype demonstrates the feasibility of a tension based, magnetically driven implantation system and opens the door to a wide variety of new minimally invasive probe materials and configurations.

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1. Introduction

Since their first use 50 years ago (Strumwasser, 1958), chronic implantable electrodes have become a critical tool for exploring the function of neural tissue. Chronic electrode recordings contribute to our basic understanding of how neural networks process sensory stimuli and impart motor control. These same electrodes can be used to actively stimulate regions of the brain, mimicking natural sensory input (Schmidt et al., 1996) and initiating physical responses (Moritz et al., 2008). The ability to interface directly with the brain is making it possible for the blind to see (Dobelle, 2000), deaf to hear (Otto et al., 2002), and paralyzed to move (Hochberg et al., 2006). While great strides have been made in neural interfa-

cial technologies, signal degradation due to glial scar formation still limits the functional lifespan of chronically implanted electrodes (Szarowski et al., 2003; Williams et al., 1999).

The continued presence of an implanted electrode initiates a chronic foreign body reaction. Over time, the body isolates the probe; activated astrocytes and microglia encapsulate the electrode, effectively displacing local neurons, hindering diffusion, and increasing interfacial impedance (Polikov et al., 2005). Micromotion of the probe relative to surrounding tissue exacerbates this phenomenon, enhancing scar tissue formation in high stress areas (Gilletti and Muthuswamy, 2006; Lee et al., 2004; Subbaroyan et al., 2005).

One of the fundamental problems with current electrodes is the dichotomy between the materials properties needed for implantation and the properties required to minimize long-term neural trauma. Current microelectrodes must be stiff enough to retain their shape under the compressive load needed to drive them through neural tissue. Silicon microelectrodes possess the required stiffness but sacrifice flexibility. The large mismatch between the modulus of silicon-based probes (~172 GPa) (Weppelmann et al., 1993) and the brain (~0.1 MPa) (Hirakawa et al., 1981) has been identified as a contributing factor in glial scar formation and subsequent long-term signal degradation (Lee et al., 2005; Subbaroyan et al., 2005). Efforts have been made to reduce scarring by developing coatings which

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improve neural cell attachment and growth (He and Bellamkonda, 2005; Ignatius et al., 1998) and prevent astrocyte adhesion (Singh et al., 2003). Such measures fail to address the severe mechanical mismatch between neural tissue and the probes themselves.

A number of relatively flexible probes have been developed to reduce micromotion-induced tissue strain. Such devices are typically fabricated from polymers such as polyimide (Cheung et al., 2007; Mercanzini et al., 2008; Metz et al., 2004; O'Brien et al., 2001; Rousche et al., 2001; Stieglitz and Gross, 2002; Takeuchi et al., 2004) using Bio-MEMS techniques, and have elastic moduli of ~ 2 GPa (Rousche et al., 2001; Takeuchi et al., 2004). Implantation using traditional techniques is difficult due to their increased flexibility. Rousche et al. (2001) addressed this issue by creating an incision in the pia prior to insertion (Rousche et al., 2001). Sharp et al. (2006) created a shape memory probe that slowly exerts force on neural tissue, allowing for gradual plastic deformation of the tissues surrounding the device (Sharp et al., 2006). Richardson-Burns et al. developed a technique for polymerizing flexible conductive poly(3,4-ethylenedioxythiophene) in direct contact with living neural tissue (Richardson-Burns et al., 2007a; Richardson-Burns et al., 2007b). Other schemes include stiffening polymeric probes by means of fluid filled channels (Takeuchi et al., 2005; Ziegler et al., 2006), bonding stiff metallic elements to the polymer (Takeuchi et al., 2004), or using ridged structures to penetrate the tissue, leaving the flexible probe behind after withdrawal (O'Brien et al., 2001). Despite using soft materials, the majority of such devices must compromise flexibility for the ability to exert enough compressive force for implantation.

This paper describes a novel system for implanting a new type of flexible neural probe. An induced magnetic field is used to accelerate a magnetic tip tethered to a flexible length of conductive wire into neural tissue. The tether, pulled by the magnetic tip, is held in tension during acceleration, preventing buckling. Preliminary in vivo results demonstrate that such probes can be used to discriminate single-unit activity, although the current dimensions of the

recording tip somewhat limits spatial resolution. We hypothesize that further refinement of this prototype system will make it possible to safely implant soft, flexible probes in an accurate and precise manner into deep brain structures for chronic neural recording. The concurrent use of flexible electrode materials is hypothesized to reduce micromotion-induced stress, thereby lessening the severity of chronic glial scarring.

2. Materials and methods

2.1. Inductive coil design

An inductive coil was constructed by wrapping 6 rows of 16 gauge copper heavy armored poly-thermaleze magnet wire (Belden) around a polyethylene spindle (Fig. 1). The coil was measured to have an inductance of $110 \mu\text{H}$ and resistance of $118 \text{ m}\Omega$. A small toroidal magnet was placed at one end of the coil to position the electrode prior to insertion. A 9 in. borosilicate glass Pasteur pipette (VWR) was inserted into the coil to act as the ejection tube (Fig. 2).

2.2. Electrode design

A smooth conical steel tip (maximum diameter 0.6 mm; length 4 mm; mass $7.2 \pm 0.4 \text{ mg}$ ($n=7$)) was soldered to a 5 cm length of 38-gauge copper heavy armored poly-thermaleze magnet wire (Belden). A section of stripped 30-gauge copper magnet wire was attached to the opposite end of the lead to serve as a recording interface (Fig. 1). The assembled insertion system is depicted in Fig. 2. The total mass of the probe (steel tip and wire) was measured to be $14.8 \pm 0.6 \text{ mg}$. To decrease the recording surface area, the steel probe tip was dip coated in a polystyrene solution. Following the coating procedure, the end of coated probe tip was scraped with a scalpel blade to expose the recording surface.

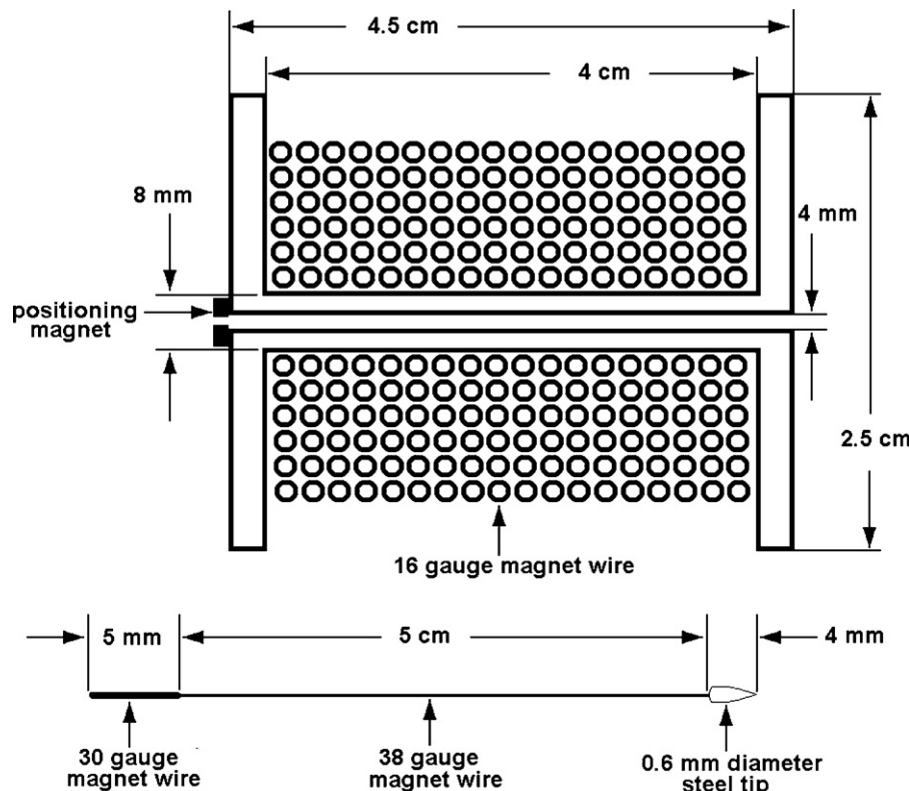


Fig. 1. Cross-section schematic of prototype inductive coil and electrode.

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