Flexible Wirelessly Powered Implantable Device

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***Abstract*— Brain implantable devices have various limitations in terms of size, power, biocompatibility and mechanical properties that need to be addressed. This paper presents a neural implant that is powered wirelessly using a flexible biocompatible antenna. This delivers power to an LED at the end of the shaft to provide a highly efficient demonstration. The proposed design in this study combines mechanical properties and practicality given the numerous constraints of this implant typology. We have applied a modular structure approach to the design of this device considering three modules of antenna, conditioner circuit and shank. The implant was fabricated using a flexible substrate of Polyimide and encapsulated by PDMS for chronic implantation. In addition, finite element method COMSOL Multiphysics simulation of mechanical forces acting on the implant and shank was carried out to validate a viable shank conformation-encapsulation combination that will safely work under operational stress with a satisfactory margin of safety.**

Keywords*— *Biocompatibility, Flexible, Modular Design, Wireless Power Transmission.

I. INTRODUCTION

Rapid progress in neuroscience has been achieved by the emerging technologies in flexible electronics and microelectronics [1]. With the ability to insert small, flexible and wirelessly powered devices in the brain, interesting and promising diagnostics and therapeutic capabilities could be generated in the near future. Neural implants are one such class of biomedical devices utilising the advancements in microelectronics for the treatment of neurological disorders, such as epilepsy or Parkinson's. In general, a neural probe is required for interfacing with the neural tissue to deliver the therapeutic effect into the region of interest. While several peripheral neural implants such as muscle microstimulators, cochlear implants and retinal implants have gained FDA approval, central nervous system (CNS) neural implants are still under development due to various design challenges [2].

The main problems in conventional neural implants are mechanical mismatch, size, shape, stiffness, and battery-based power supply. All these problems pose difficult challenges for long term integration for the implants. Among the long-term failure modes, nearly half of them are mechanical failures [3], and up to 80% of those are of connector problems. Mechanical mismatch is known to cause chronic inflammation, and therefore long-term

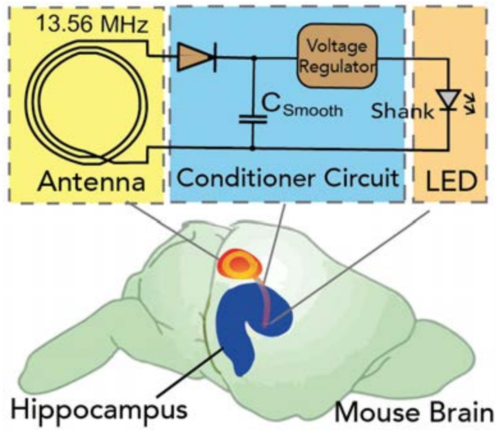


Fig. 1. Conceptual model of the flexible neural implant device in mouse brain.

incompatibility in the brain tissue due to trauma induced by micromotion, foreign body reaction due to shape, size and material surface and finally, rejection of the implant [3]. In order to minimise this, an ideal implant should be designed to have similar density and stiffness with the native tissue, should be as small and thin as possible and without tethering. Additionally, biocompatibility of design materials is essential in reducing foreign body response to the implant, ensuring long term viability. The most commonly used materials are Polydimethylsiloxane (PDMS), Parylene-C and Polyimide, due to their favourable mechanical, electrical and chemical characteristics in the use of neural prosthetics [2]. Using finite element analysis, the implantable neural device can be optimised with respect to material choice and parameters such as encapsulation thickness and dimensions.

Studies have reported the advantages of Wireless Power Transmission (WPT) systems against conventional battery-based systems. There are three main WPT approaches: ultrasonic, photovoltaic, and electromagnetic coupling. The suitability of each is dependent on the application. Given the restrictions of every implantation case, the choice of the WPT crucially determines the long-term viability. The trend in implantology is to mitigate the negative effects of battery-based or movement-based generators, which are typically large and biologically incompatible [3]. It is also essential to achieve a small size and reduced stiffness for better biocompatibility [4, 5].

Ultrasonic WPT is based on the piezoelectric effect, where a piezoelectric energy harvester implanted inside the body is resonated in order to generate an induced voltage.

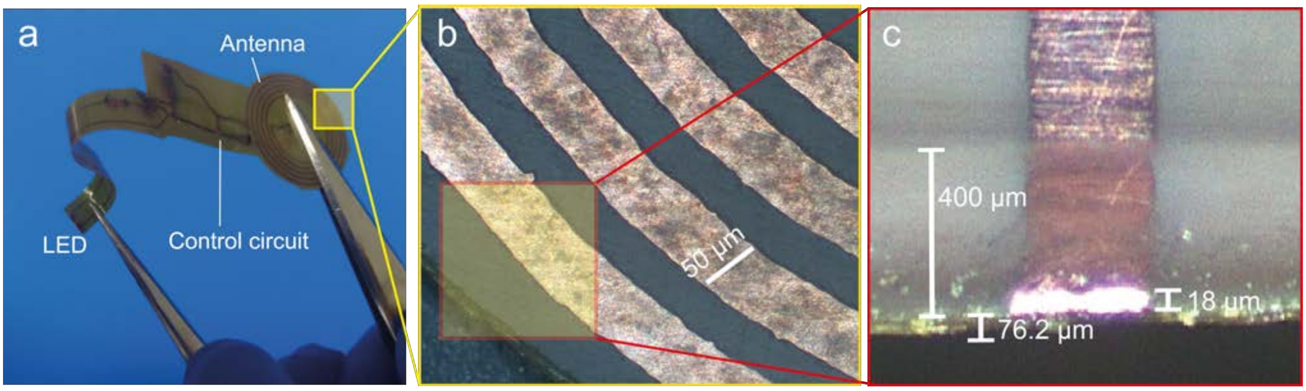


Fig. 2. (a) View of the fabricated implant in Pyralux AP8030R using a positive etching removing the copper from the non-circuit areas through mechanical etching and ferrite chloride. (b) Zoom to the WPT module of the encapsulated device illustrating the PDMS coating and the layer model of the device. (c) Zoom in cross section of the device with polyimide, copper and PDMS layers of different thickness.

However, this technology is not sufficiently developed and so it is not commonly used for flexible implantable devices [6].

On the other hand, photovoltaic wireless power harvesting is being considered as a novel approach for many biomedical applications [7,8]. This technique is based on the conversion of light into electricity using photovoltaic cells. Although the advantage of this technology is ease of focusing the source to the receiver, transmission distance is far more inefficient compared to the other WPT technologies [9,10,11,12].

Although the emerging WPT technologies seem to be attractive, electromagnetic coupling is the most commonly used approach at the moment. It is still the most reliable in terms of tissue compatibility and power transmission efficiency [13]. Many research groups are trying to miniaturise the WPT device by employing inductive coupling while using industrial, scientific and medical (ISM) radio band frequencies [13,14,15].

In this work, we are proposing a wirelessly powered, flexible and modular implant design approach, consisting of (i) an antenna for wireless power transmission, (ii) a conditioner circuit and (iii) a shank with LED for demonstration. Adopting a modular approach means that the shank could be redesigned in future, introducing electrodes for electrical stimulation. This design approach is illustrated in the Fig. 1. Finite element simulations were run along the device fabrication to validate the model and to analyse the mechanical stress during operation. Finally, we have optimised the encapsulation layer by material and thickness for fabrication.

II. METHODOLOGY

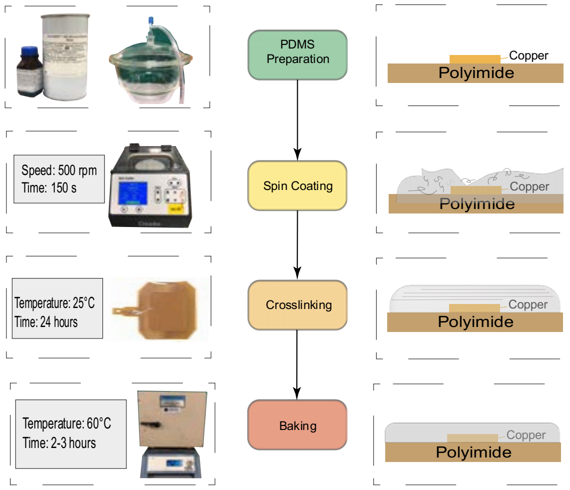
A. Implant design and Layout

The long-term viability of the neural implant is dependent on both geometry and size. Therefore, for optimal biomechanical compliance, a small and suitable shape must be achieved. A small prototype design has been proposed, to the dimensions of 40 mm length by 23 mm width for the probe base and 40 mm by 4 mm for the shank. The choice of substrate and encapsulation materials is essential in order to match the electromechanical requirements for ideal operation. Polyimide (Dupont AP 8535R) was chosen for its well-proven biocompatibility, electrochemical inertness and flexibility [16]. The polyimide thickness is 76.2 µm and 18 µm for the copper. Polyimide, PDMS and Parylene-C were also considered for the encapsulation.

The proposed design was simulated in *COMSOL Multiphysics* to study the 90° bend implantation case [1], where the shank is inserted into the brain and the probe base is attached to the scalp. Here, Von Mises stress over the probe is reported with a specific interest in the shank-probe junction (knee), where the stress is considered to be the highest. Two key steps have been taken to optimise the design: *(i)* encapsulation material optimisation between Polyimide, PDMS and Parylene-C were simulated and *(ii)* varying PDMS encapsulation thicknesses were simulated. After material optimisation, an interpolation model for finding the optimal stress-thickness relationship for the device fabrication is also presented.

B. Electronic Design

The electronic design is divided into three sections: antenna, conditioner circuit and shank. In Fig. 1, the conceptual diagram of this device is illustrated. The WPT system consists of a circular coil designed with an outer and inner diameter of 22.3 mm and 12.3 mm, respectively. Although dimensions of the designed antenna are large in comparison to the mouse brain, the modular approach provides the option of individual design of the WPT system exclusively. Both the wire width and wire spacing were fixed at 50 µm, as shown in Fig. 2a. The coil was designed with 5 turns obtaining a theoretical inductance of 0.992 µH and therefore, according to the impedance matching of a parallel RLC resonant circuit typology. A matching capacitance of 138 pF was used to tune the receiver coil at 13.56 MHz. The conditioner circuit consists of a half-wave rectifier using a Schottky diode (NSR05F40NXT5G) with a low forward voltage drop of maximum 500 mV connected to a smoothing capacitor (2.2 nF) in parallel. After the rectification and smoothing of the signal, a linear Low-Dropout (LDO) Voltage Regulator (NCP161) was included which can supply



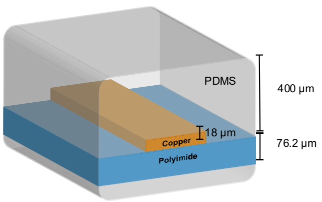


Fig. 3. Encapsulation protocol divided in 4 steps: PDMS preparation, Spin coating, crosslinking and Baking.

450 mA to the shank. In this work, an LED for testing the performance of the circuit was placed at the end of the shank (4 cm) with two copper traces, one for input and other for ground. Depending on the length of the copper traces, the load on the shank as well as the electromagnetic compatibility can be modified. In Fig. 2(a), the fabricated device in the conformal form is shown.

C. Encapsulation and biocompatibility

There are several studies using polymers for neural implant encapsulation [17]. However, the most widely used material is PDMS (polydimethylsiloxane). Its flexibility, resistance to biodegradation and high biocompatibility make it suitable for encapsulation [17]. The whole process of encapsulation is illustrated in Fig. 3, and it comprises four steps: PDMS preparation, spin coating, crosslinking and baking. To cure the PDMS, Slygard 184 and curing agent was mixed to a ratio of 10:1 in a temporary container at room temperature for 5 minutes. The solution was degasified using vacuum desiccator to remove air bubbles formed during the mixing process. PDMS was left to cure at room temperature for 10 minutes. An Ossila spin coater was used to spin coat cured PDMS on the polyimide substrate. Since the uniformity of the encapsulation thickness is directly proportional to spin time, different spin speeds and times were tested to achieve an optimal, thin coating. For our circuit, we spun 1.25 g of PDMS at 500 rpm for 150 s. For crosslinking, PDMS deposited substrate was left at room temperature for 24 h and then it was dried in the oven at 60°C for 3-4 h. The formula for calculating the theoretical thickness is given below:

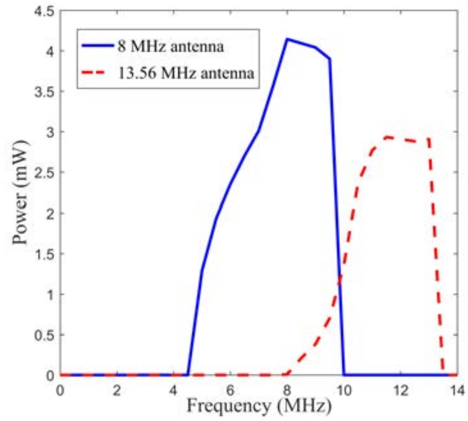


Fig. 4. Power performance on the 8 MHz and 13.56 MHz frequency antennas.

$$h = \left(\frac{3\mu}{4\rho}\right)\frac{1}{w\sqrt{t}} \qquad (1)$$

where, h is the thickness of PDMS, w is the spin speed while t is the spin time, μ is the viscosity of the cured PDMS and ρ the density of the PDMS [18]. Given (1), the encapsulation thickness obtained was 400 µm which can be further reduced to micron using higher spin speed and time.

D. Fabrication Methods

In this study, Dupont Pyralux AP 8535R has been used as substrate. Therefore, the excess copper from both layers of the polyimide have been etched away and only copper traces remain. Typical methods of etching are suitable for this substrate. In this study, ferric chloride was used as an etchant and typical mechanical process was followed. The cross section of the device is illustrated in Fig. 2c, and consists of four main layers: (i) polyimide sheet of thickness 76.2 µm, (ii) copper trace of thickness of 18 µm, (iii) electronic components comprised of epoxy resins and, (iv) encapsulation layer of PDMS with thickness of 400 µm.

III. Simulation and Experimental Results

Fig. 4 compares the performance of the 8 MHz and 13.56 MHz antennas. A signal generator was used to generate a sine wave of 10 V of amplitude and a frequency sweep from 0 to 14 MHz was carried out. The results obtained prove that the performance of the device around the operation frequency reach a maximum power of 4 mW in the case of the 8 MHz and 3 mW in the case of the 13.56 MHz antenna. The power in the shank presents the same value within the bandwidth of the WPT. This experiment also successfully proves the modularity of the proposed device as two different antennas and different transmission frequencies were used with the same implant. As expected, the electrical performance of the device remains similar even with different antennas. Additionally, the successful electrical operation of the device is demonstrated in different bending configurations. The results for the probe model in the bending implantation case at 90° in *COMSOL Multiphysics* are given in Fig. 5. To mimic

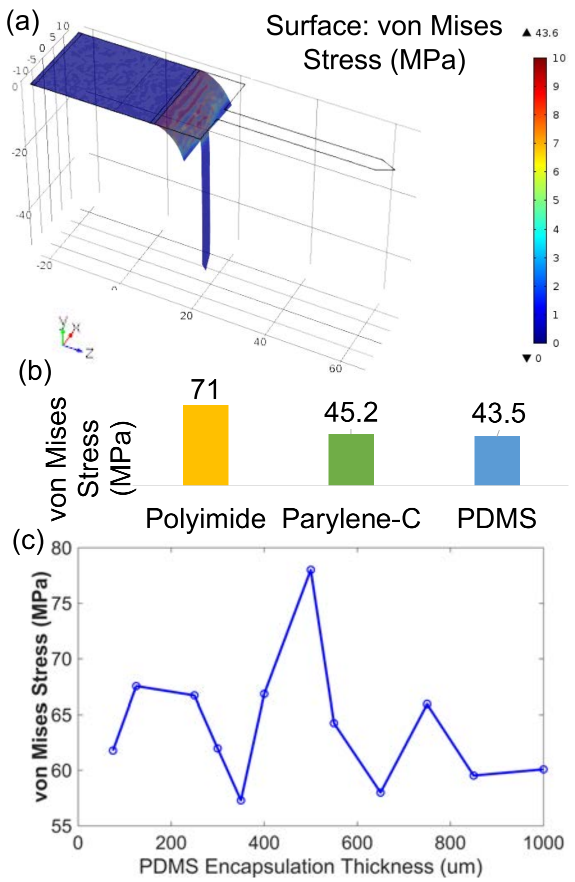


Fig. 5. (a) Simulation in COMSOL Multiphysics for the implantation case and simulation model. (b) Material simulation results. (c) PDMS Thickness simulation.

the implantation scenario, loading force is applied on the shank to bend it, to achieve the conformation that represents the implantation case. For material choice of encapsulation, in Fig. 5(b), Parylene-C, Polyimide and PDMS encapsulations have been simulated. Polyimide shows double the von Mises stress over the surface compared to the PDMS. Since PDMS has shown the lowest stress, it has been confirmed as the choice of encapsulation, and further analysed for optimum thickness. PDMS encapsulation optimisation model by layer thickness versus von Mises stress are given in Fig. 5(c) showing that the optimal encapsulation thickness is around 350µm, which is close to the fabricated thickness of 400µm.

IV. Conclusion

In this work, the design and fabrication of a flexible and wirelessly powered neural implant utilising an innovative modular design is presented. Regarding this novel approach, advantages over conventional devices in terms of adaptability have been empirically demonstrated. An interchangeable WPT module operating at 8 MHz and 13.56 MHz has been implemented proving the advantage of the modularity approach. The device was successfully tested in different frequencies and bending conformations. Additionally, an interpolation model for finding the optimal stress-thickness relationship for the device fabrication has been presented, running a mechanical simulation in *COMSOL Multiphysics*. Finally, the implant has been successfully encapsulated using 400 µm of PDMS using the optimisation model.

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