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 an implantable device/probe that powered wirelessly using a flexible biocompatible antenna. A Wireless Power Transfer system for powering a LED was devised to provide a high-efficient demonstration. The proposed design in this study combines optimal size and practicality given the mechanical 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 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 in the tissue due to trauma induced by

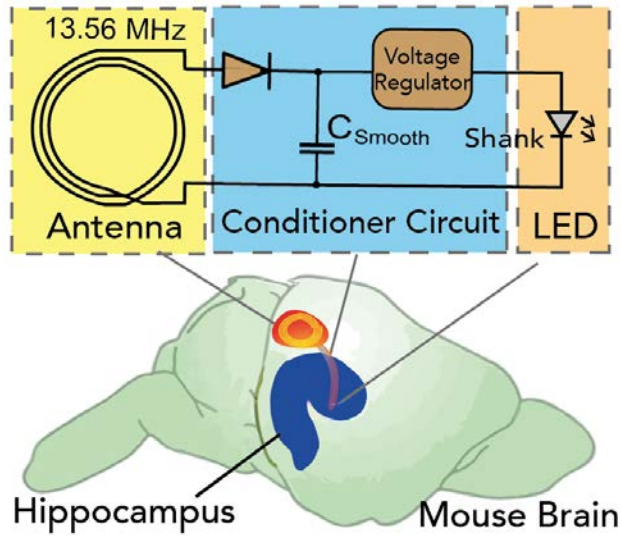

Fig. 1. Conceptual model of the flexible neural implant device in mouse brain.

micromotion. 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 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.

Some studies have reported the advantages of Wireless Power Transmission (WPT) systems against conventional battery-based powering. There are three main WPT approaches; ultrasonic photovoltaic and electromagnetic coupling. Depending on the application it is necessary to determine which one is more suitable. Given the restrictions of every implantation case, the choice of the WPT crucially determines the long-term viability. Furthermore, for brain implants it is essential to achieve a small size and 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. However, this technology is not very developed yet and so, it is not commonly used for flexible implantable devices [6].

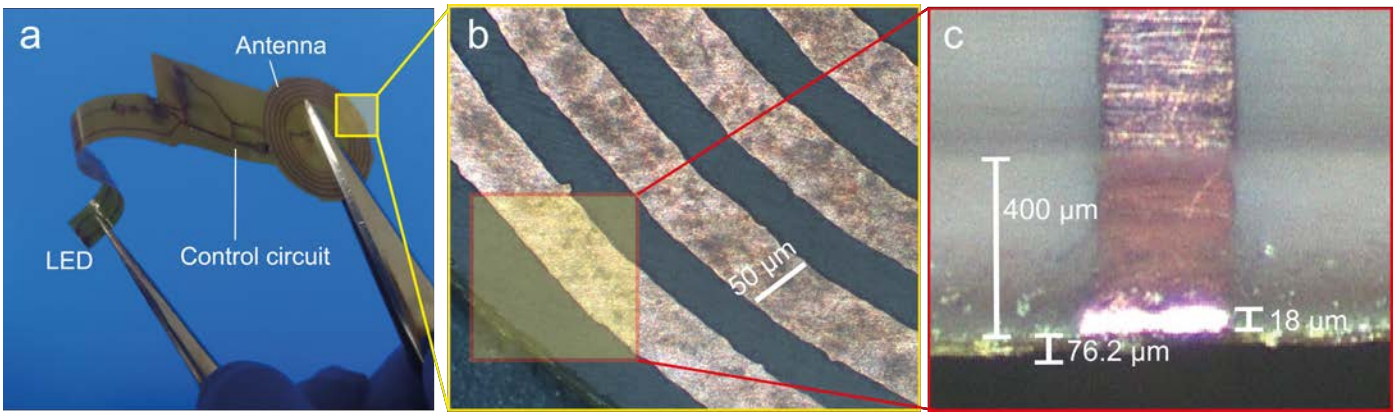

Fig. 2. (a) Transversal view of the antenna's copper traces over the polyimide. (b) Cross section of the encapsulated antenna module illustrating the PDMS coating and the layer model of the device. (c) Zoom in cross section of the device with polyimide, copper and PDMS layers of different thickness.

On the other hand, photovoltaic wireless power harvesting is being considered as a novel approach for many biomedical applications [7,8]. Photovoltaic wireless power harvesting is based on the conversion of light into electricity by using photovoltaic cells. Although the advantage of this technology is ease of focusing the source to the receiver, transmission distance is far 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]. Furthermore, inductively wirelessly powered implants are currently in the spotlight of many research groups, who are trying to miniaturise the size further while using industrial, scientific and medical (ISM) radio bands 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. 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

Geometry and size are two factors that highly affects the long-term implantation of the neural implant. 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 40mm by 4mm for the shank. The choice of substrate and encapsulation materials is essential in order to match the electromechanical requirements for ideal operation. Among all the materials available for flexible substrates for implants, 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. Regarding the encapsulation, Polyimide, PDMS and Parylene-C were considered.

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. For the optimisation of the design, two steps have been taken: *(i)* encapsulation material optimisation between Polyimide, PDMS and Parylene-C were simulated and *(ii)* varying PDMS encapsulation thicknesses have been simulated. After material optimisation, an interpolation model for finding the optimal stress-thickness relationship for the device fabrication has also been 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. 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 450 mA to the shank. In this work, an LED for testing the performance of the circuit was used, 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

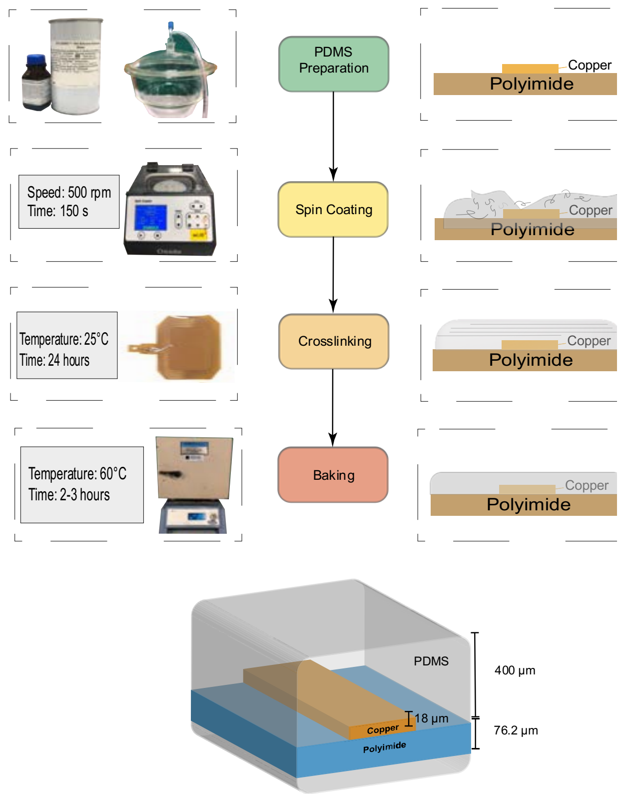

Fig. 3. Encapsulation protocol divided in 4 steps: PDMS preparation, Spin coating, crosslinking and Baking.

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). It's flexibility, resistance to biodegradation and high biocompatibility make its use suitable for encapsulation [17]. In implants, encapsulation long-term integration without showing signs of degradation while keeping its biocompatibility intact. The whole process of encapsulation is illustrated in Fig. 3, and it is comprised of four steps: PDMS preparation, spin coating, crosslinking and baking. Slygard 184 has been used for encapsulation. 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. Ossila spin coater was used to spin coat cured PDMS on the polyimide substrate. To obtain a thin and uniform PDMS encapsulation layer, different spin speeds and times were tested. Since the uniformity of the encapsulation thickness is directly proportional to spin time, adequate times have been applied. For our circuit, we spin 1.25g of PDMS at 500 rpms for 150 secs. For crosslinking, PDMS deposited substrate was left at room temperature for 24 hrs and then it

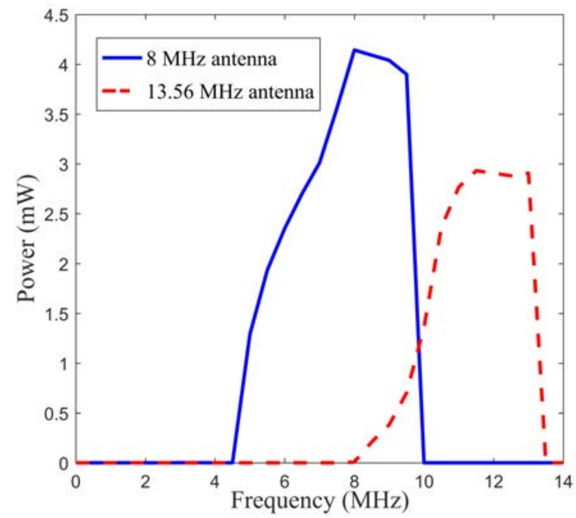

Fig. 4. Power performance on the 8 MHz and 13.56 MHz frequency antennas.

was dried in the oven at 60ºC for 3-4 hrs. The formula for calculating the theoretical thickness is given below:

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

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 the density of the PDMS [18]. Given Eq. 1, the encapsulation thickness obtained is 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, by etching method the copper from both layers of the polyimide have been removed 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. 2(a-c), and is comprised 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

In the Fig. 4, the comparison between the performance of the 8 MHz and 13.56 MHz antennas is given. The output power of the implant in the terminal shank was measured with a frequency sweep from 0 to 14 MHz The results obtained suggest that the performance of the device around the operation frequency is stable obtaining 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. This experiment also successfuly proves the modularity of the proposed device. As expected, the electrical performance of the device remains similar even with different antennas. Additionally, the successful electrical operation of the device has been demonstrated in different bending configurations. The results for the probe model in bending implantation case at 90° in *COMSOL*

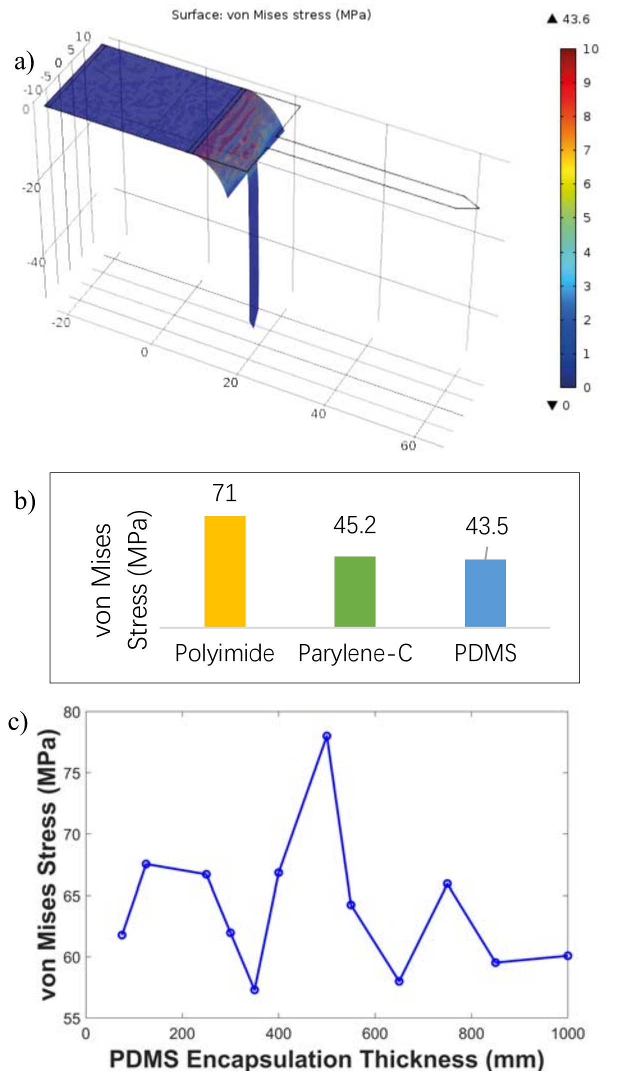

Fig. 5. (a) Simulation in COMSOL Multiphysics for the implantation case and simulation model. (b) Material simulation results. (c) PDMS Thickness simulation.

Multiphysics are given in Fig. 5. To mimic 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 using 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.

References

[1] H. Heidari, N. Wacker and R. Dahiya. "Bending induced electrical response variations in ultra-thin flexible chips and device modelling", *Appl. Phys. Rev.*, vol. 4, no. 3, May 2017.

[2] M. Salam, J. L. Perez-Velazquez, R. Genov, "Seizure suppression efficacy of closed-loop versus open-loop deep brain stimulation in a rodent model of epilepsy", IEEE Trans. Neural Syst. Rehabil. Eng., vol. 24, no 6, p. 710-719, 2015.

[3], D. Prodanov and J. Delbeke. "Mechanical and biological interactions of implants with the brain and their impact on implant design", *Front. neurosci.*, vol. 10, no. 11, Feb. 2016.

[4] R. Das, and H. Yoo. "A multiband antenna associating wireless monitoring and nonleaky wireless power transfer system for biomedical implants", *IEEE Trans. Microw. Theory Tech.*, vol. 65, no. 7, pp. 2485-2495, Jul 2017.

[5] L., Yu-Po, et al. "A battery-less, implantable neuro-electronic interface for studying the mechanisms of deep brain stimulation in rat models", *IEEE Trans. Biomed. Circuits Syst.*, vol. 10, no 1, pp. 98-112, Feb. 2016.

[6] A. Ibrahim, M. Meng, M. Kiani, "A comprehensive comparative study on inductive and ultrasonic wireless power transmission to biomedical implants". IEEE Sensors, vol. 18, no 9, p. 3813-3826, 2018.

[7] K. Oo Htet, *et al.*, "Energy-Efficient Start-up Power Management for Batteryless Biomedical Implant Devices". In: IEEE Int. Conference on Electronics, Circuits and Systems (ICECS), pp. 597-600, Dec. 2018.

[8] J. Zhao, R. Ghannam, Q. Abbasi, M. Imran and H. Heidari, (2018) Simulation of Photovoltaic Cells in Implantable Application. In: IEEE Sensors 2018 Conference, Oct. 2018.

[9] K. Oo Htet, *et al.*, "Power management using photovoltaic cells for implantable devices", IEEE Access, vol. 6, pp. 42156 – 42164, 2018.

[10] J. Zhao, K. Oo Htet, R. Ghannam, M. Imran. H. and Heidari, "Modelling of Implantable Photovoltaic Cell based on Human Skin Types". In: Conf. PhD Research in Microelectronics Electronics (PRIME), Jul 2019,

[11] J. Zhao *et al.,* "Design, test and optimization of inductive coupled coils for implantable biomedical devices". J. Low Power Electron., vol. 15, no. 1, pp. 76-86., 2019.

[12] M. Meng, M. Kiani, "Design and optimization of ultrasonic wireless power transmission links for millimeter-sized biomedical implants", IEEE Trans. Biomed. Circuits Syst, vol. 11, no 1, p. 98-107, 2016.

[13] S. Y. R. Hui, W. Zhong and C. K. Lee. "A critical review of recent progress in mid-range wireless power transfer", *IEEE Trans Power Electron*, vol. 29, no. 9, pp. 4500-4511, Mar. 2013.

[14] M., Kate L., et al. "Wirelessly powered, fully internal optogenetics for brain, spinal and peripheral circuits in mice", *Nat. meth.*, vol. 12, no. 10, p. 969-974, Aug 2015.

[15] P. Gutruf et al. "Fully implantable optoelectronic systems for battery-free, multimodal operation in neuroscience research", *Nat. Electron.*, vol. 1, pp. 652-660, Dec. 2018.

[16] T., Adrian et al. "Polymeric biomaterials for medical implants and devices". *ACS Biomater. Sci. Eng,* vol. 2, no. 4, pp. 454-472, Feb. 2016,

[17] C. Hassler, T. Boretius and T. Stieglitz. "Polymers for neural implants", *J. Polym. Sci. B Polym. Phys.*, vol 49, no. 1, p. 18-33, Jan. 2011.

[18] R. D. Rivers." Design, Fabrication, and Characterization of an Electrostatically Actuated Microfluidic Valve". *MSc. in Materials Engineering. California Polytechnic State University*, Jun. 2010.