Design of a Flexible Receiver Module for Implantable Wireless Power Transfer (WPT) Applications

M. Sinclair, D. Biswas, T. Le, J. Hyde, I. Mahbub
Department of Electrical Engineering, University of North Texas

L. Chang
Department of Biomedical Engineering, University of North Texas

Y. Hao
Department of Mechanical Engineering, Northwestern Polytechnical University

***Abstract*— For biomedical applications, implantable devices require an external or internal power source to operate. The restricting wires and battery limitations emphasize the need for wireless power transfer (WPT) system for the implants. In this paper, a flexible receiver module is presented for WPT applications using Polyethylene Naphthalate (PEN). The WPT system achieve a maximum power transfer efficiency (PTE) of 0.15% at 5 mm distance through air at 4 MHz operating frequency. The specific absorption rate (SAR) for the proposed WPT system is evaluated through a 6-layer tissue model.**

Introduction

With the advancement of medical technologies, various implants have been designed for diagnosing and treating illnesses. With the increasing demand for implants, the power requirements for these devices need to be improved [1]. Batteries, a traditional power source, have their drawbacks. Recharging or replacing the batteries may require surgery, which can lead to complications [2]. WPT is a viable alternative to batteries to deliver power efficiently to the implants. The principle of WPT is to transfer power from an external transmitting coil (TX) to an internal receiving coil (RX) without the use of wires. A WPT system for implants using a rigid substrate is presented in [3]. The WPT operates at 2.8 MHz and achieves a peak power transfer efficiency of 30% and 4.3% at 5 mm and 12 mm distance through air respectively. Previously, a miniaturized WPT system is fabricated on a rigid FR4 substrate using copper traces and it operates at 7.15 MHz [4]. The WPT system achieves a maximum efficiency of 4.1% at a 5 mm distance through air.

Devices, as in [3] and [4], with rigid substrates cannot accommodate to the contours of tissue while being used as implants. However, devices with flexible substrates can be made smaller and implanted in a wider variety of locations than devices with rigid substrates [5]. Additionally, any antenna or device implanted inside the human body is required to have a specific absorption rate (SAR) below 1.67 W/Kg as per IEEE regulations [6].

This paper evaluates the Power Transfer Efficiency (PTE) and SAR of a WPT system across a 6-layer tissue model using High-Frequency Structure Simulator (HFSS). For the proposed WPT system, a flexible substrate is used in the design of the RX while a rigid substrate is used for the TX. The Power Transfer Efficiency (PTE) of the fabricated system is also evaluated through measurements.

Design and Fabrication of WPT System

The presented WPT system consists of a TX module and a RX module as shown in Fig. 1.The TX is fabricated on a FR4 substrate since the TX is not required to be implanted. The dimension of the two-turn TX planer spiral coil is chosen as 20 × 25 mm^2 with 1 mm wide copper traces (Fig. 2(a)). The RX is designed to be placed under or above the skull layer in the head as shown in Fig. 2(b). The RX module is designed using Polyethylene Naphthalate (PEN), a flexible substrate with a permittivity (ε_r) of 4.2 and a bending modulus of 2.2 GPa. The RX module has a dimension of 1.9 × 2.8 cm^2 and our previous work shows the implementation of it on FR4 substrate [7]. The RX has a 9.8 × 9.8 mm^2 planar spiral with 10 turns, as shown in Fig. 2(c). The metal lift-off process is used to fabricate the RX module traces. First, the photoresist is patterned on PEN substrate as the sacrifice layer. Then, gold is sputtered on the PEN with the thickness of about 100 nm for the conducting layer. A thicker layer of gold reduces the resistance of the RX module and improve the PTE of the WPT system. Finally, the photoresist is lifted off using acetone to form the planer spiral coil pattern. The RX module is composed of commercially available off-the-shelf (COTS) components: Schottky diode (CUS08F30 by Toshiba Semiconductor), voltage regulator (TPS6120DRCT by Texas Instruments), and surface-mounted inductors, resistors, and capacitors. The receiver module has a 1.625 kΩ load resistor to emulate the load and the power requirement.

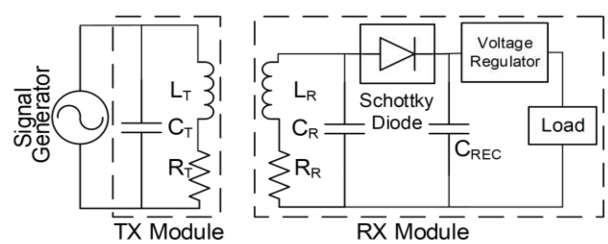


Fig. 1. Block diagram of a WPT System.

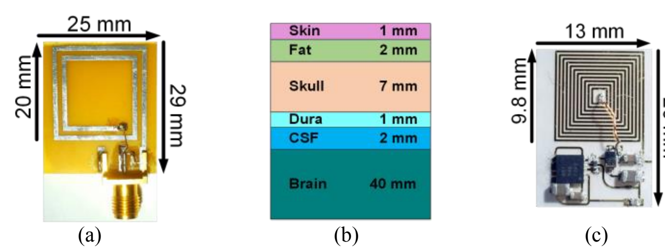
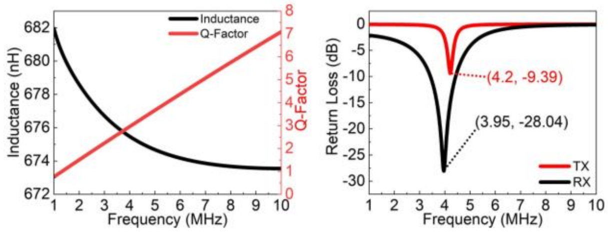
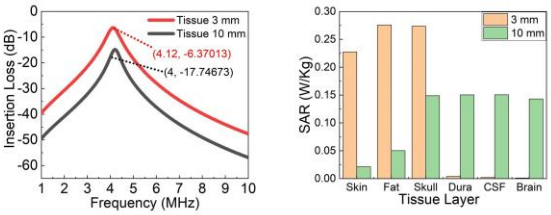


Fig. 2. (a) TX on FR4, (b) Six-layer tissue model, (c) RX on PEN.



(a) (b)

Fig. 3. (a) Inductance and Q-Factor of RX coil, (b) Simulated Return loss (S_{11}) of the TX and RX modules.



(a) (b)

Fig. 4. (a) Insertion Loss S(2,1) and (b) SAR at different tissue depths.

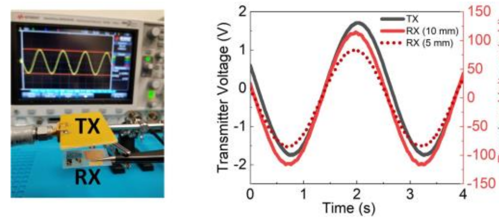
DESIGN OPTIMIZATION AND VALIDATION

High-Frequency Structure Simulator (HFSS) is used to simulate the WPT system. The desired operation frequency, *f*, is chosen to be approximately 4 MHz due to the low absorption of power through the tissue media at that frequency. At 4 MHz, the RX coil has the simulated *Q*-factor of 3 and an inductance of 676 nH as shown in Fig. 3(a). The TX coil has a *Q*-factor of 37 and an inductance of 38 nH at the same frequency. To resonate at 4 MHz, the RX and TX require a resonating capacitor of 2200 pF and 11 nF respectively.

The simulated return loss (S_{11}) of the RX and TX with the resonating capacitors are shown in Fig. 3(b). The TX return loss is -9.83 dB at 4.2 MHz. The RX return loss is -28.3 dB at 3.95 MHz. Both the TX and RX resonate at approximately 4 MHz which validates the chosen operating frequency. In HFSS, the SAR is simulated using the six-layer tissue model, composed of skin, fat, skull, dura, CSF, and the brain [6]. The transmitted power is chosen to be 15 dBm for this simulation. The insertion loss (S_{21}) of the WPT system through the six-layer tissue and 1.5 mm of air is -17.7 dB for 10mm of tissue distance and -7.21 dB for 3 mm of tissue distance (Fig. 4(a)). The simulated PTE for 3mm and 10 mm tissue depth is 20% and 2.4% respectively. The maximum SAR for each tissue layer is shown in Fig. 4 (b). The proposed WPT system shows SAR values in the individual layers that are well below the maximum 1.67 W/Kg limit, which indicates that the internal RX can be safely implanted inside the human head.

MEASURING PTE THROUGH AIR

The PTE of the fabricated WPT system is measured through the air for experimental validation. The transmitter power is generated using a Keysight EXG analog signal generator N5171B and interfaced to the TX module with an SMA connector. The transmitted power is 15 dBm at 4 MHz frequency. The TX and RX are positioned 10 mm apart which is assumed to be the maximum distance between the TX and the implanted RX. The RX output voltage is measured across the spiral coil and the resonating capacitor using a Keysight infiniiVision DSOX3014A oscilloscope. The measured PTE is the received power divided by the transmitted power as a percentage.



(a) (b)

Fig. 5. (a) Test set-up at 10 mm, (b) Measured TX and RX voltages.

Fig. 5(a) shows the test set-up for the measurement. The measured peak to peak voltage across the RX resonating capacitor is 230.2 mV_{pp} and 168.8 mV_{pp} for 5 mm and 10 mm distance respectively (Fig. 5(b)). Using equation (1), the PTE is calculated as 0.15% at 5 mm and 0.08% at 10 mm distances. The drop in efficiency could be due to the high-resistivity of the conductive epoxy that is used for assembling the components on the PEN substrate. Also, the inductance of the RX coil could be slightly different than the simulated inductance, which might cause the shift in the resonance frequency of the RX and thus the drop in the PTE.

CONCLUSION

The WPT system presented in this paper has been designed using an RX on a flexible PEN substrate and a TX on rigid FR4 substrate. The preliminary results of the flexible substrate based RX module shows promising aspects for more compact and biocompatible implants. A future challenge of the flexible substrates is the characterization of misalignment and displacement due to the bending that may impact the PTE.

REFERENCES

[1] T. Sun, X. Xie and Z. Wang, "Design challenges of the wireless power transfer for medical microsystems," *2013 IEEE International Wireless Symposium (IWS)*, Beijing, 2013, pp. 1–4.

[2] X. Wei and J, Liu, "Power sources and electrical recharging strategies for implantable medical devices", *Frontiers of Energy and Power Engineering* in China, 2(1), 2008, pp.1–13.

[3] R. Wu, S. Raju, M. Chan, J. K. O. Sin and C. P. Yue, "Silicon-Embedded Receiving Coil for High-Efficiency Wireless Power Transfer to Implantable Biomedical ICs," in *IEEE Electron Device Letters*, vol. 34, no. 1, pp. 9–11, Jan. 2013.

[4] D. K. Biswas, M. Sinclair, J. Hyde and I. Mahbub, "An NFC (near-field communication) based wireless power transfer system design with miniaturized receiver coil for optogenetic implants," *2018 Texas Symposium on Wireless and Microwave Circuits and Systems (WMCS)*, Waco, TX, 2018, pp. 1–5.

[5] G. Xiao, Z. Zhang, H. Fukutani, Y. Tao and S. Lang, "Improving the Q-Factor of Printed HF RFID Loop Antennas on Flexible Substrates by Condensing the Microstructures of Conductors," in *IEEE Journal of Radio Frequency Identification*, vol. 2, no. 2, pp. 111–116, June 2018.

[6] L. K. Ragha and M. S. Bhatia, "Evaluation of SAR Reduction in Implanted Spherical Head Model Using RF Shield," *2009 Annual IEEE India Conference*, Gujarat, 2009, pp. 1–4.

[7] D. K. Biswas, N. T. Tasneem, J. Hyde, M. Sinclair and I. Mahbub, "Miniaturized Wireless Power Transfer Module Design for Brain Optoelectronic Implant," *2018 IEEE International Microwave Biomedical Conference (IMBioC)*, Philadelphia, PA, 2018, pp. 163–165.