Implantable RF Power Converter for Small Animal *In Vivo* Biological Monitoring

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Abstract—**A miniature, long-term, implantable radio frequency (RF) power converter for freely moving small animal *in vivo* biological monitoring is proposed. An environment consisting of a laboratory mouse inside a cage is used for a prototype monitoring system design. By employing an inductive coupling network, a prototype implant device with a dimension of approximately 6 mm x 6 mm x 1 mm and a weight of 100 mg including medical-grade silicone coating can wirelessly receive an input RF power from an array of external coils positioned underneath the cage. Each coil is designed to be 5 cm x 5 cm, comparable to a typical mouse size for minimizing power coupling variation. The received AC voltage is further rectified by a half-wave rectifier to supply DC current to a 3 kΩ load resistance, representing a typical bio-implant microsystem loading. The proposed RF power converter was implanted in the peritoneal cavity of a laboratory mouse for performance evaluation. With a 5-turn external coil loop separated from a 30-turn internal coil by 1 cm distance and centered to each other, an optimal voltage gain of 3.5 can be achieved with a 10 MHz operating frequency to provide a maximum rectified output DC voltage of 21 V. The DC voltage varies at different animal tilting angles and positions with a minimum voltage of 4 V at 60º tilting angle near the corner of the external coil. This variation can be further minimized by overlapping the external loops layout. A voltage regulator can also be designed to provide a stable supply for an overall bio-implant system.**

Keywords—**Biomedical implant, *In vivo* monitoring, RF powering system, Inductive coupling**

I. INTRODUCTION

DNA sequencing of laboratory mice together with *in vivo* real time biological information, such as blood pressure, temperature, activity, and bio-potential signals, is ultimately crucial for systems biology research, genetic function discovery, and new treatments development for diseases. A miniature, long-term, reliable bio-sensing implant network is highly desirable to capture a dynamic behavior of a biological system. Existing commercial implant devices are inadequate to achieve the objectives due to their large size and weight, causing severe post-implant trauma and data distortion, and limited functionalities. The implants are typically dominated by (rechargeable) batteries or large coil loops for remote powering and data telemetry. By employing low-power integrated electronics technology and miniature RF-powering system, the implant size and weight can be significantly reduced [1-4]. However, providing a sufficient and stable energy to power the implant system inside a freely moving animal is challenging. In this paper, an implantable RF power converter for small animal *in vivo* biological monitoring is proposed. The prototype system employs an inductive coupling technique to convert an RF power to a DC voltage with a 3 kΩ load resistance, representing a typical bio-implant microsystem loading.

II. IMPLANT SYSTEM ARCHITECTURE

Figure 1 presents the overall wireless implant microsystem architecture with RF powering and data telemetry capabilities.

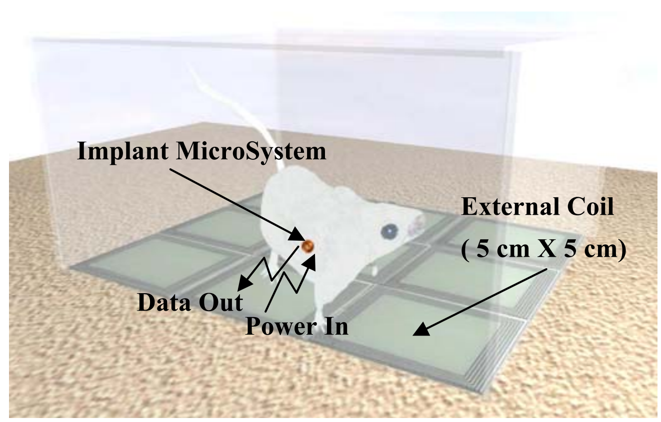

Figure 1. Implant Microsystem Architecture

The implant device would receive an input RF power from an array of external coils positioned underneath the cage. The received signal is further rectified and regulated to supply a stable DC power to the implant microsystem. To minimize a large voltage coupling variation and low coupling efficiency, each external coil is designed to be 5 cm x 5 cm, comparable to a typical mouse size. The external arrays can potentially be adaptively controlled depending on the animal position.

A simplified remote RF powering system consisting of an inductive coupling network and a half-wave rectifier is illustrated in Figure 2.

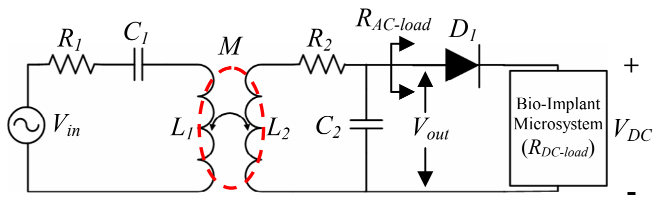

Figure 2. Simplified Remote RF Powering System

A tuned series resonator consisting of L_1 and C_1 is driven by an external RF power source, V_{in}. A resistor, R_1, represents the total series resistance associated with the resonator including the output resistance of the power source. In the implant side, a parallel resonator, consisting of L_2 and C_2 with a total loop resistance of R_2, is used to receive the RF power from the external circuitry. The received RF signal is then rectified by a diode, D_1, to produce a DC supply voltage. In the prototype design, a $R_{DC\text{-}load}$ of 3 kΩ is employed to represent a typical bio-implant microsystem loading. When both coils are properly tuned to the same resonant frequency, ω, a voltage gain can be expressed as [5]

$$V_{out} / V_{in} = \frac{\omega^2 L_2 M}{(R_1 R_2 + (\omega M)^2 + R_1 (\omega L_2)^2 / R_{AC-load})}, \quad (1)$$

where M represents the mutual inductance between L_1 and L_2, and $R_{AC\text{-}load}$ is the equivalent AC resistive loading presented to the parallel resonator by the half-wave rectifier, which can be determined as [6-7]

$$R_{AC-load} = R_{DC-load} / 2 . \quad (2)$$

In order to achieve an efficient power conversion system, the voltage gain across the tuned resonator network needs to be maximized. From Equation (1), all the parameters, L_1, L_2, M, R_1, and R_2, are a function of coil geometry, coupling distance, and operating frequency. Due to the size of the external coil loop being fixed as 5 cm x 5 cm with a normal separation of 1 cm from the internal coil loop, L_1, M, and R_1 can be varied by changing the coil turn numbers. Moreover, due to the implantable device size constraint and coil self resonant frequency, the internal coil loop is limited to thirty turns. Thus, the inductive coupling system can be extensively characterized to obtain an optimal condition for achieving a maximum voltage gain. The measured voltage gain versus operating frequency for various external coil loop turn numbers is presented in Figure 3 (internal coil and external coils being separated by 1 cm and centered to each other), which closely matches the analytical results from Equation (1).

It can be seen that an optimal voltage gain of 3.5 can be achieved at 10 MHz operating frequency with a five-turn external coil loop exhibiting an inductance value of 2.9 µH and high-frequency resistance of 4 Ω. The thirty-turn internal coil loop provides an inductance value of 3.4 µH and high-frequency resistance of 16 Ω accordingly. The two loops exhibit a mutual inductance of 50 nH and self resonant frequencies well above 50 MHz. Experimental data also shows that an external loop with an increased turn numbers would result in a reduced voltage gain due to self resonance effects.

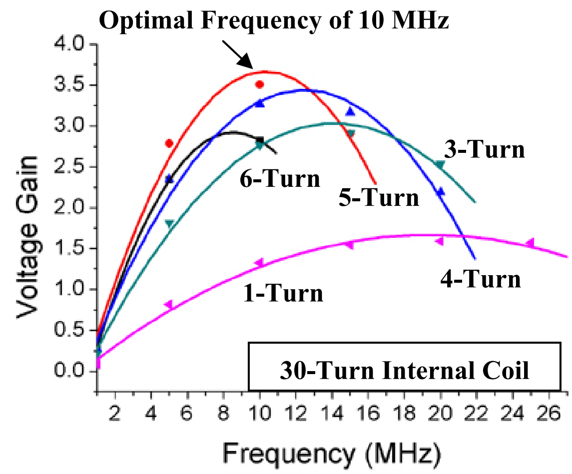

Figure 3. Measured Voltage Gain vs. Operating Frequency with Different External Coil Turn Numbers

Figure 4 shows a schematic of the external circuitry consisting of an RF power source driving a class-E amplifier [8]. The five-turn external coil loop, L_1, with a 1 mm line width and 0.5 mm spacing is embedded in a printed circuit board (PCB) as a part of the class-E amplifier layout with a tuned capacitor, C_1, RF coil choke, L_{choke}, and shunt capacitor, C_{shunt}. A 10 MHz clock signal is applied to drive the switching transistor, M_1, for generating an RF power signal. From SPICE simulation, an RF signal with a 4.4 V peak to peak amplitude can be achieved from a 6 V DC supply with C_{shunt} of 10 nF, which can provide an adequate power for the bio-implant microsystem.

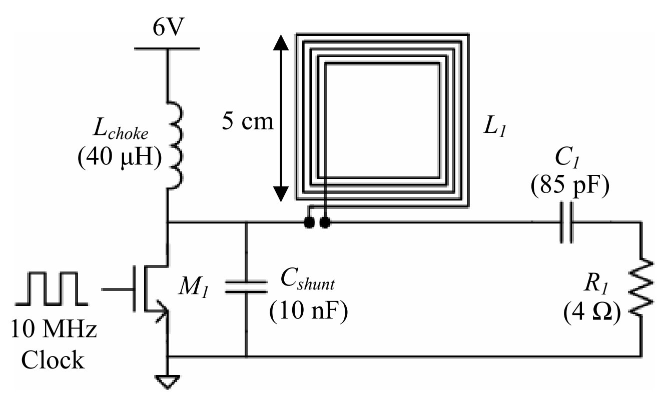

Figure 4. External Circuitry Employing Class-E Amplifier

Figure 5 presents a picture of a medical-grade silicone-coated prototype design consisting of a hand-wound internal inductive coil loop and rectifier circuit with a dimension of approximately 6 mm x 6 mm x 1 mm and a weight of 100 mg. The internal loop made by a plan enamel magnet wire 34 AWG is attached on a 130 µm-thick copper PCB with a surface mounted diode, D_1, tuned capacitor, C_2, and 3 kΩ loading resistance. The device is then covered by a biomedical-grade elastomer material (MDX4-4210) from Dow Corning Company. A center void (~3 mm x 3 mm) is intended for future CMOS chip placement, on which the rectifier will be integrated with other sensing electronics.

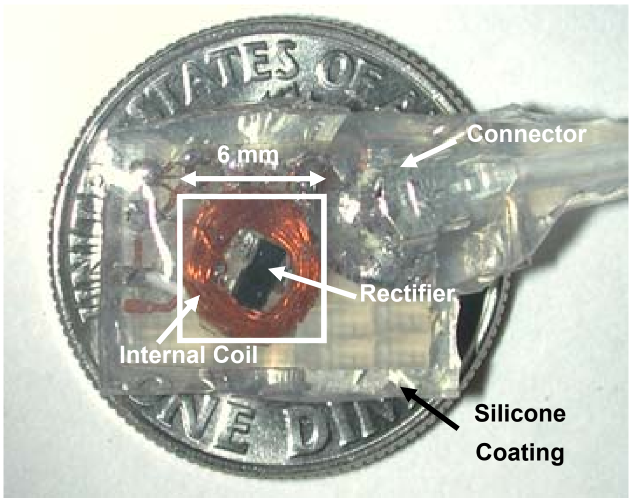

Figure 5. Prototype Implantable Device

The achieved size and weight would represent at least one order of magnitude improvement compared to existing commercial devices. A further weight reduction can be expected by avoiding an excessive silicone coating on the implantable device.

III. Measurement Results

The prototype implant setup is illustrated in Figure 6. A one inch incision was made in the peritoneal cavity of a laboratory mouse, in which the device was implanted parallel to the frontal plane. The incision was closed using 5-0 dexon suture. The wound was closed leaving an opening just large enough for the silicone coated wire lead to protrude.

Figure 7 shows the characterization experimental setup. It can be seen that typically, when a mouse walks or sleeps, the implant device is approximately 1 cm apart from the external coil in a parallel orientation. Therefore, the first characterization experiment was conducted based on this general behavior. Table 1 shows the measured rectified output voltage, V_{DC}, when the implant device is characterized outside and inside the mouse body at different positions with respect to the external coil loop.

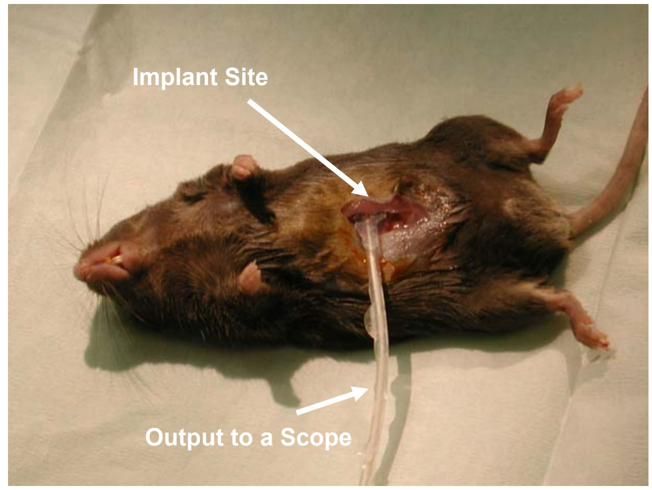

Figure 6. Prototype RF Powering Device Implanted in a Laboratory Mouse

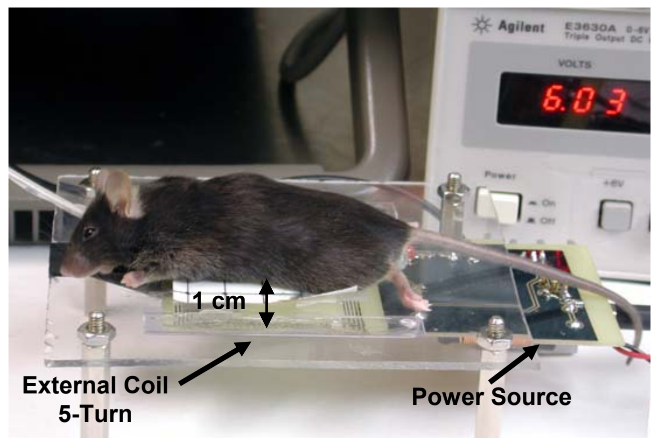

Figure 7. Experimental Setup with the Implant Device Placed Parallel to the External Coil Loop

Position	Outside Animal	Inside animal
Center	6.5 V	6.4 V
Edge	3.1 V	2.95 V
Corner	2.48 V	2.39 V

Table 1. DC Output Voltage (V_{DC}) with the Device Tested Outside and Inside the Mouse Body

Based on the optimal voltage gain illustrated in Figure 3, the maximum V_{DC} is expected to be 6.9V, which matches well with the measurement value of 6.5 V. The experimental data also indicates that the skin absorption to an RF signal at 10 MHz has a negligible effect. In order to characterize the actual behavior of a freely moving mouse, the second experiment was conducted with different animal tilting angles as illustrated in Figure 8 with measurement results presented in Table 2. Experimental data indicates that the voltage generally decreases as the tilting angle increases because of a reduced flux linkage from the external coil loop to the internal coil loop. Due to the circulating magnetic flux lines from the external coil loop, it can be shown that the flux lines near the coil edge exhibit an increased leaning angles compared to the coil center, thus resulting in an enhanced power coupling efficiency when the internal coil is tilted at a matching angle. From the current setup, the reduced output voltage can be further improved by overlapping the external loops layout. A voltage regulator following the rectifier will also be necessary to achieve a sufficient and stable power supply independent of the mouse location and tilting angle [5].

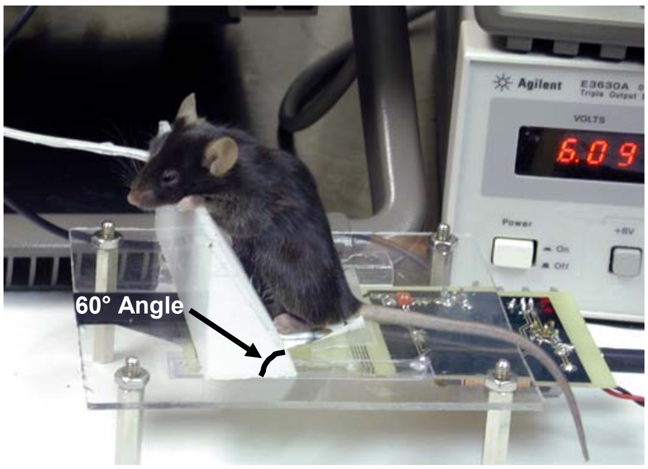

Figure 8. Experimental Setup with the Implant Device Tilted by 60° with Respect to the External Coil

Tilting Angle	Center (outside/ inside mouse)	Edge (outside/ inside mouse)	Corner (outside/ inside mouse)
30°	2.3V/ 2.4V	3.4V/ 3.3V	1.9V/ 2.1V
45°	2.2V/ 2.1V	3.1V/ 3.2V	1.8V/ 1.9V
60°	1.4V/ 1.3V	3.2V/ 3.3V	1.0V/ 1.13V

Table 2. Output Voltage at Different Tilting Angles

With a further optimization of the class-E power amplifier by properly adjusting its tuning network, *in vitro* experiments were conducted to demonstrate an increase in coupling voltage amplitude, achieving a maximum and minimum voltage of 21 V and 4 V, respectively.

IV. Conclusion

A miniature, long-term, implantable RF power converter for freely moving small animal *in vivo* biological monitoring is proposed. The prototype implant device can wirelessly receive an RF power from an external coil positioned underneath the cage. An optimal voltage gain of 3.5 can be achieved with a 10 MHz operating frequency to supply a 3 kΩ resistive loading with a rectified DC output voltage of 21 V when a five-turn external coil loop is separated from a thirty-turn internal coil by 1 cm distance and centered to each other. The rectified output voltage varies at different animal tilting angles and positions with a minimum voltage of 4 V at 60º tilting angle near the corner of the external coil. This variation can be further minimized by overlapping the external loops layout. A voltage regulator can also be designed to provide a stable supply for an overall bio-implant system.

Acknowledgment

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