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Wireless Powering and Data Telemetry for Biomedical Implants

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Abstract—**Wireless powering and data telemetry techniques for two biomedical implant studies based on (1) wireless *in vivo* EMG sensor for intelligent prosthetic control and (2) adaptively RF powered implantable bio-sensing microsystem for real-time genetically engineered mice monitoring are presented. Inductive-coupling-based RF powering and passive data telemetry is effective for wireless *in vivo* EMG sensing, where the internal and external RF coils are positioned with a small separation distance and fixed orientation. Adaptively controlled RF powering and active data transmission are critical for mobile implant application such as real-time physiological monitoring of untethered laboratory animals. Animal implant studies have been successfully completed to demonstrate the wireless and batteryless *in vivo* sensing capabilities.**

I. Introduction

Wireless powering and data telemetry have been widely employed as a critical enabling technology for advanced biomedical implant applications, where it is impractical or impossible to connect an implant device directly to an external power source or a processing unit [1-8]. Inductive coupling allows an external RF power source to couple its power from a tuned series LC resonator to a parallel LC tank, which is designed as a part of the implant system. The coupled RF signal will be rectified and regulated to provide a stable DC power supply for the implant. Inductively coupling also enables short-distance passive data telemetry based on amplitude shift keying (ASK) and phase shift keying (PSK) modulations with zero DC power dissipation [9]. Active data transmitter is employed for achieving a large telemetry distance, however, with a penalty of an increased system power dissipation [10]. Various data modulation schemes such as on-off keying (OOK), ASK, and frequency shift keying (FSK) have been demonstrated for active wireless data transmission. Research results also show that OOK and ASK can be susceptible to amplitude variation of received signals due to implant device movement such as untethered laboratory animal real-time monitoring, whereas FSK modulation is much more robust to mitigate these effects [10]. Therefore, adequate RF powering and data telemetry techniques need to be carefully selected for different biomedical implant applications.

In this paper, we present two biomedical implant cases studies based on (1) wireless *in vivo* EMG sensor for intelligent prosthetic control and (2) adaptively RF powered implantable bio-sensing microsystem for real-time genetically engineered mice monitoring. The two applications are based on different operating conditions, thus calling for different RF powering as well as data telemetry techniques.

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II. Wireless *In Vivo* EMG Sensor

Myoelectric signals are critical natural control sources for achieving an enhanced biomimetic performance of powered prosthetic lower limbs. An implantable wireless EMG recording capability is highly desirable for mitigating a number of undesirable aspects associated with the current implementation of myoelectically-controlled prostheses. The proposed *in vivo* EMG sensing microsystem is designed to stream real-time EMG data when the implantable coil is brought into a close alignment with the external power transmitting coil as depicted in Figure 1, where an external transceiver is mounted in a prosthetic socket with a wireless link coupled to an EMG sensor implanted in a residual limb.

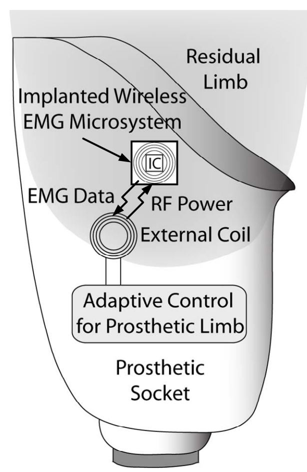


Fig. 1. Prosthetic system architecture

Figure 2 presents the overall wireless microsystem design architecture consisting of an inductively coupled wireless powering and data telemetry link and integrated CMOS sensing and telemetry electronics. A Class-E RF power amplifier drives a series-tuned resonant spiral coil located inside the prosthetic socket. A miniature implantable parallel-tuned resonant coil is positioned coaxially with respect to the external coil with a separation distance up to 2 cm, corresponding to a typical gap size for the proposed application. The inductively coupled RF power is rectified and regulated to generate a stable on-chip power source for the implant electronics. A passive digital phase-shift keying (PSK) data telemetry, which shares the same wireless inductive link as the powering system, is selected due to its zero DC power dissipation.

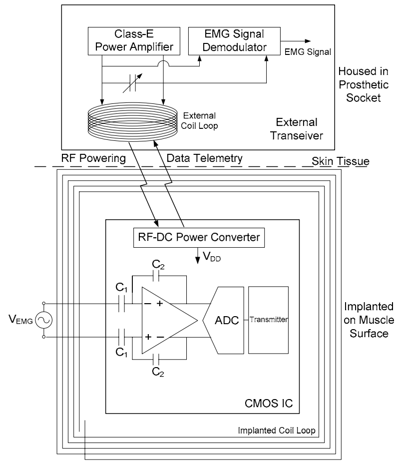


Fig. 2. Wireless EMG sensing microsystem architecture

Figure 3 presents the detailed wireless powering and data telemetry architecture. An external Class-E power amplifier operating at an optimal frequency of 8 MHz drives a 24-gage series-tuned resonant spiral coil (L_1) exhibiting a 2-cm diameter, 10-turn, and 2 µH inductance with a series resistance of approximately 1 Ω to inductively couple the RF power to the implantable 36-gage parallel-tuned resonant spiral coil (L_2) exhibiting a 6-mm diameter, 20-turn, and 3.5 µH inductance with a 6 Ω series resistance over a separation distance of 1 to 2 cm [11]. The RF power system achieves a power transfer efficiency of 8% with an inductance coupling factor of approximately 0.03. The received RF signal is then converted into a stable 2V and 2.7V DC supplies to power the implant electronics as a load with a total current consumption of 83 µA; 63 µA of which is drawn from 2V supply. The 2.7V DC supply is used to effectively switch MOSFET switches employed in the implant microsystem design.

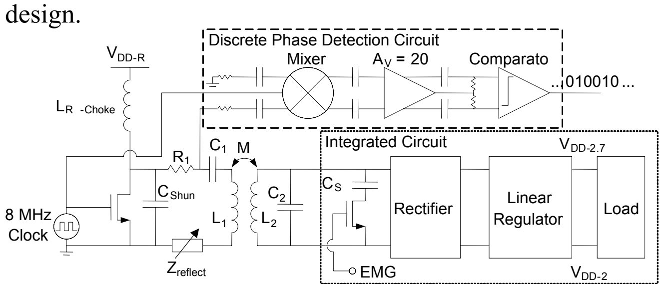


Fig. 3. Wireless powering and data telemetry architecture

The digital encoded EMG bitstream is wirelessly transmitted to an external receiver by passive digital phase shift keying over the same wireless link shared by the inductive powering system through switching a detuning capacitor, C_{sw}, in parallel with the implantable secondary LC tank, thus causing a complex impedance, $Z_{reflect}$, to be reflected to the external primary tank [9, 11], as shown in Figure 3. The detuning capacitor is selected to ensure a minimum degradation (less than 10%) in the RF coupling efficiency. The EMG data can be demodulated by using an external phase detection circuit and interfaced with a PC for data logging and analysis. Figure 4 shows a photo of the prototype wireless EMG sensing microsystem. A 2.2 mm x 2.2 mm ASIC containing all required electronic functions is enclosed in the middle of a 6-mm diameter RF coil with two discrete capacitors for a compact design. The overall microsystem characterization shows that the implantable EMG sensing electronics can be wirelessly powered by an external RF source and send digital PSK data to an external receiver over the same inductive link. Figure 5 presents the Manchester-encoded transmitted data and the corresponding recevied data with a bit-error-rate (BER) well below 10^{-6} [11]. Laboratory rat implant study further demonstrates the wireless sensing capability of real-time EMG signals as shown in Figure 6 [12].

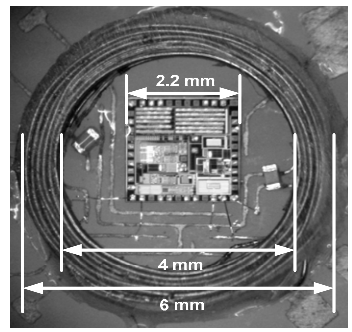


Fig. 4. Prototype wireless EMG sensing microsystem

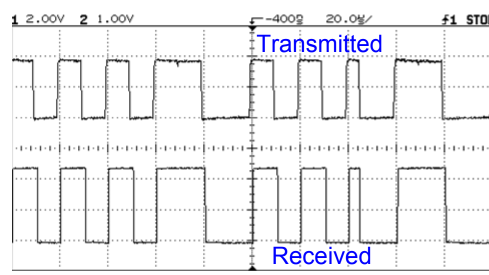


Fig. 5. Manchester-encoded PSK data telemetry

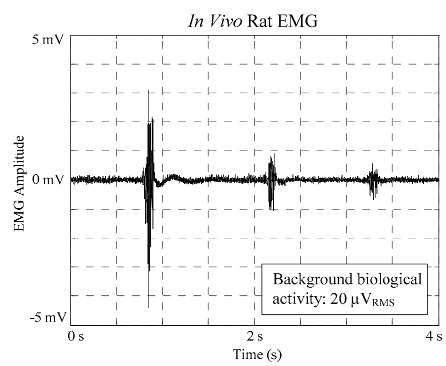


Fig. 6. Wirelessly received *in vivo* EMG activity from laboratory rat

III. Adaptively RF Powered Implantable Bio-Sensing Microsystem For Real-Time Genetically Engineered Mice Monitoring

DNA sequencing of laboratory mice together with *in vivo* real-time biological information, such as blood pressure, core body temperature and bio-potential signals, is ultimately crucial for advanced biomedical and genetic research to identify genetic variation susceptibility to diseases and to potentially develop new treatment methods for similar human diseases. A wireless, batteryless, and implantable multi-channel bio-sensing microsystem is highly desirable to capture real-time biological information from a "free" roaming animal housed in its cage as depicted in Figure 7.

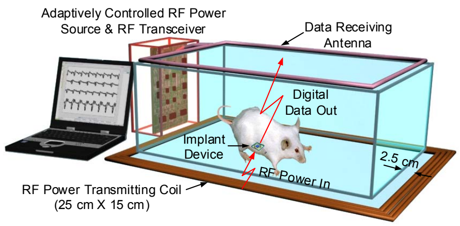
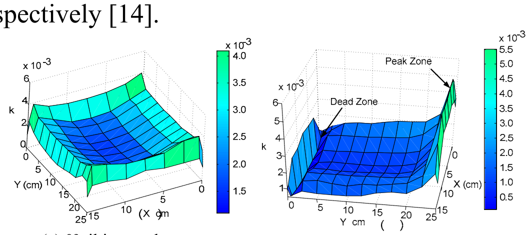


Fig. 7. Wireless batteryless implantable bio-sensing system architecture

In conventional RF powering, the relative position of the internal coil is fixed with respect to the external coil. However, in the case of an untethered laboratory animal implant, the internal coil is inside a freely moving laboratory animal, thus resulting in a continuously changing RF power coupling. Therefore, an adaptive RF powering is required to achieve a sufficient and stable energy to power the implant electronics in a varying magnetic field. The wireless implantable system consists of integrated bio-sensing and telemetry electronics with passive components such as RF power receiving coil, data transmitting coil, and filtering capacitors. The RF power receiving coil can be winded in a planar spiral configuration to reduce the overall system thickness for a compact design. As shown in Figure 7, an external adaptively controlled RF power source driving an RF power transmitting coil positioned underneath the cage is used to couple the RF power to the implant unit. The implant electronics first detect the incoming RF power level and then transmits this information together with other real-time biological signals to a nearby receiver. The retrieved power level data will be used to adjust the external RF power so that a desired constant implant DC power can be achieved independent of the animal's movement.

A mouse cage area of 25 cm x 15 cm is chosen for the prototype design; hence, the size of the external RF power transmitting coil. For optimally designing the adaptive RF power control system, the coupling factors at different cage locations were characterized by using the prototype RF power coils with a normal separation distance of 1 cm. An optimal operating condition was determined to use a 4-turn external coil and a 20-turn 5-mm-diameter internal coil with an operating frequency of 4 MHz [13]. The measured coupling factor (k) with an animal tilting angle of 0° and 60°, which is the worst-case tilting considered for the prototype design, are presented in Figures 8a and 8b, respectively [14].



(a) 0° tilting angle (b) 60° tilting angle

Fig. 8. Measured coupling factor with the external coil separated from the internal coil by 1 cm

The k values vary from 0 to 0.55%, where the minimum coupling (dead-zone) and maximum coupling (peak-zone) occur around the edge of the external coil. The dead-zone can cause an unreliable RF power reception, and the peak-zone can result in an excessive power coupling. Therefore, by limiting the cage size at 2.5 cm away from the external coil edge, the dead-zone and peak-zone can be excluded. With a cage size of 20 cm x 10 cm, k values are bounded from 0.078 % to 0.34 %. This operating condition is sufficient to generate an on-chip 2V DC supply with up to 1 mA current driving capability for the integrated electronics over the chosen cage size with an animal tilting angle of up to 60°. Still, the large k variation can result in a significant RF power variation. Without the adaptive feedback, calculation shows that the implant would require a constant external RF power of 18.5 W for an animal being located in the small coupling area. Once moving to a higher coupling region, an excessive received RF power can result in an excessively large DC voltage as well as ripple to fail the implant electronics. However, with the adaptive feedback, the external power source is expected to dissipate 4.7 W and 20 W for the best-case and worst-case power coupling, accordingly. The received RF power level due to varying coupling factor can be detected by an on-chip RF power level sensing circuit and fed back to adjust the transmitted RF power strength as depicted in Figure 9. Due to the weak inductive coupling, passive data telemetry is inadequate for a reliable data transmission. Therefore, an active transmitter was chosen at 433 MHz with FSK modulation scheme exhibiting a frequency deviation of 210 kHz.

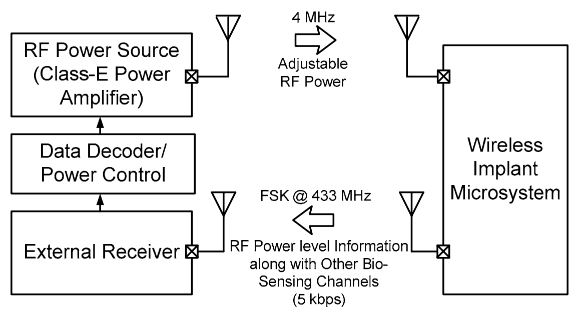


Fig. 9. Adaptive RF powering system architecture

Figure 10 qualitatively illustrates the operation of the adaptive RF power control system.

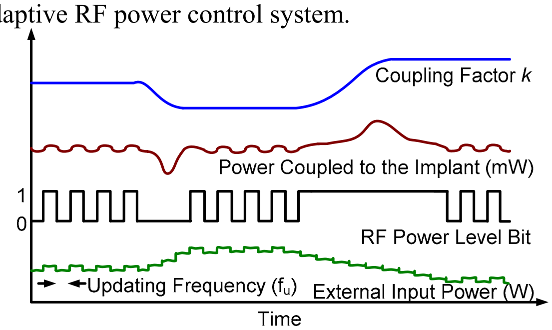


Fig. 10. Timing diagram of adaptive powering system

When k decreases, the power coupled to the implant temporarily drops, causing the RF power level bit to remain low. The adaptive controlling program steps up the external input power to regain the coupled power in the implant system to a desired level. In this prototype, the transmitting RF power is controlled by adjusting the supply voltage of a Class-E amplifier implemented as the external power source. When k increases, the external input power is stepped down to a proper level resulting in a lower external power consumption, thus a reduced RF input power. When k is constant (mouse not moving), the RF power level data bit alternates between a high and low state, causing an average constant power coupled into the implant. The coupled RF power will be rectified and regulated to provide a stable on-chip power source to supply the implant microelectronics.

Figure 11 show a prototype wireless, batteryleess and implantable blood pressure sensing microsystem for laboratory mice real-time monitoring with a 5-mm diameter circular coil for adaptive RF powering [10].

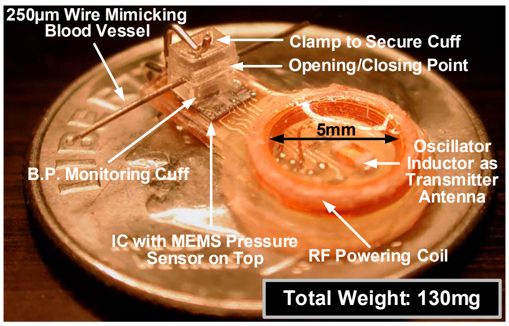


Fig. 11. Adaptively RF powered blood pressure monitoring microsystem

The blood pressure sensing microsystem was implanted in laboratory mice for *in vivo* characterization under a wireless and batteryless condition as shown in Figure 12.

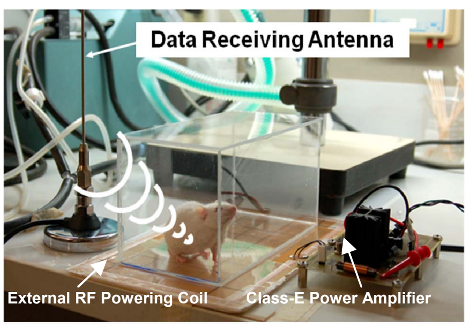


Fig. 12. Wireless *in vivo* characterization under adaptive RF powering

Aorta artery was selected for the ease of an implant procedure. The wirelessly recorded blood pressure waveform after animal recovery is shown in Figure 13.

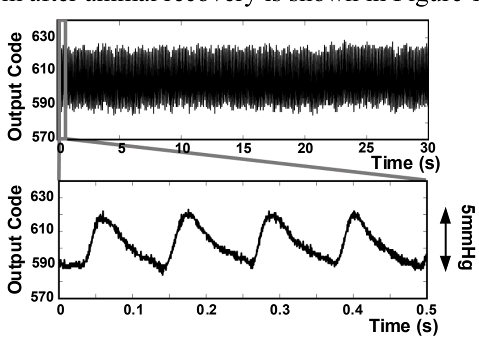


Fig. 13. Wirelessly received *in vivo* mouse blood pressure waveform

The measurement results demonstrate that the prototype microsystem can capture real-time high-fidelity blood pressure information under adaptive RF powering.

IV. Conclusion

Inductive-coupling-based RF powering and passive data telemetry is effective for biomedical implant applications, where the internal and external RF coils are positioned with a small separation distance and fixed orientation. Large telemetry distance will require an active data transmitter for reliable signal reception. Adaptively controlled RF powering is critical for mobile implant application such as *in vivo* real-time physiological monitoring of untethered laboratory animals.

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