Wireless Implantable EMG Sensor for Powered Prosthesis Control

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Abstract

This paper presents a wireless, subfascially implantable electromyogram (EMG) sensing microsystem for intelligent myoelectric control of powered prostheses. The implantable system consists of a custom-designed ASIC, an RF telemetry coil, and two Pt-Ir epimysial EMG electrodes, and is capable of wirelessly transmitting digitized EMG data to an external telemeter mounted in a prosthetic socket. The prototype microsystem is powered by a near-field inductive link operating at 8 MHz with 10% DC power transfer efficiency. On-chip rectification and regulation produce stable 2 V and 2.7 V supplies with a DC current driving capability up to 100 μA. The EMG electrodes are interfaced with a differential capacitively-coupled amplifier with 38 dB closed-loop gain, 1 kHz bandwidth, and $78\ \mathrm{nV}/\sqrt{Hz}$ input-referred noise floor. The amplified EMG signal is then digitized on chip by using an 11-bit algorithmic ADC. The digital EMG data can be Manchester-coded and transmitted to the external telemeter using passive phase shift keying (PSK) modulation scheme over the same wireless link as the inductive powering system. The implantable microsystem consumes 83 μA and achieves 8.7-bit resolution when wirelessly powered by an external RF energy source. Animal implant experiments have been successfully conducted to demonstrate EMG signals from a laboratory rat hind limb can be detected and wirelessly transmitted to an external telemeter with high signal fidelity.

1. Introduction

Myoelectric signals are critical and natural control sources for enhanced biomimetic performance of powered prosthetic lower limbs [1]. Existing implementations of myoelectrically controlled prosthetic lower limbs rely on surface EMG recorded by using either electrodes in the prosthetic socket or skin patch electrodes attached to the residual limb. The former option precludes the use of a prosthetic sock, which causes uncomfortable friction between the socket and the residual limb. The latter option requires the user to prepare the skin surface and attach the electrodes and lead wires daily. Both approaches are prone to motion artifacts and unreliable connections due to perspiration. A fully implantable EMG sensor obviates these issues by measuring muscle activity directly from the muscle surface, allowing for the use of a prosthetic sock, and eliminating any additional steps when donning the prosthesis. Wireless inductively coupled systems have been implemented for many applications, where it is impractical or impossible to connect a sensor directly to processing or actuation systems [2-4]. This is especially true in biomedical implants where wires passing through the skin are prone to infection. In the battery-less microsystem presented here, a class-E RF power amplifier drives a series-tuned resonant spiral coil located outside the body. A miniature parallel-tuned resonant coil is positioned coaxially with respect to the transmitting coil with a separation distance of up to 2 cm, corresponding to the typical gap size for this application. Passive digital phase-shift keying (PSK) data telemetry, which shares the same wireless inductive link as the powering system [4], is selected for the microsystem design due to the relatively high obtainable data rate and static coupling factor of the coils. In this paper, an EMG sensing microsystem with wireless power and data telemetry capability for intelligent prosthesis control is presented. The remote powering system produces stable DC power supplies of 2 V and 2.7 V, providing 83 μA for the integrated electronics. The implantable system can transmit a single channel of 50 kbps Manchester-coded EMG data to the external transceiver over an 8 MHz RF powering link. Animal implant experiments are conducted to demonstrate EMG signals from a laboratory rat hind limb can be detected and wirelessly transmitted to an external receiver with high signal fidelity, adequate for intelligent powered prosthesis control.

2. Wireless Microsystem Design

The proposed wireless implantable EMG sensing microsystem architecture is depicted in Figure 1. The implantable unit consists of a custom-designed low-power ASIC, a miniature radio-frequency (RF) coil, and a pair of epimysial EMG sensing electrodes. The entire unit is battery-less and is operated by a RF power inductively coupled from an external power source mounted in the prosthetic socket. Digitized EMG data can be wirelessly transmitted to a receiver, also mounted in the socket, over the same wireless power link. The received EMG data is further processed and used for adaptive prosthetic limb control.

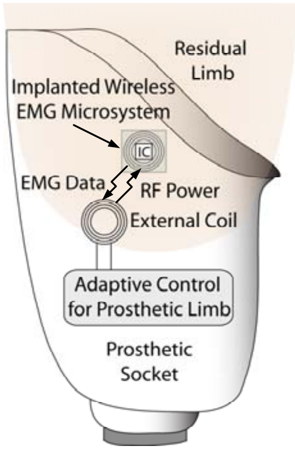

Figure 1. Wireless implantable EMG sensing architecture

Figure 2 presents the overall electronic system design architecture, which consists of wireless power and data telemetry link, integrated RE-DC power converter, and EMG sensing amplifier followed by analog-to-digital converter (ADC) and wireless data transmitter.

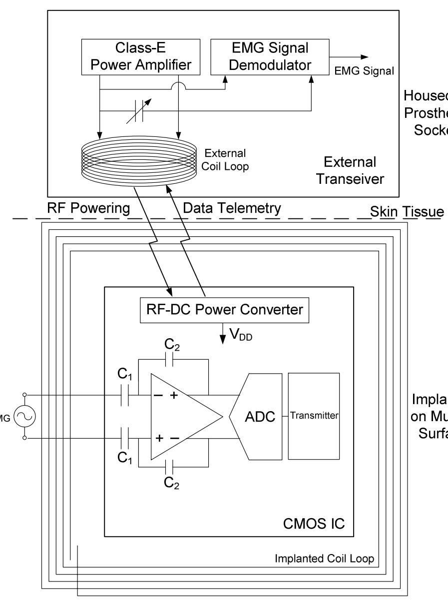

Figure 2. Electronic system design architecture

Figure 3 shows a simplified diagram of the wireless powering system. A class-E power amplifier, operating at 8 MHz, drives a set of tuned LC resonators to couple the RF energy into the implantable electronics. An on-chip voltage doubler, implemented by using diode-connected NMOSFET, M_{d1} and $M_{d2,}$ converts the RF signal to $V_{Supply-H}$ and $V_{Supply-L,}$ which are further processed by a linear regulator to generate stable DC voltages of 2V and 2.7V. The intrinsic CMOS parasitic diodes, D_1 and D_2, do not interfere with the doubler operation. The implant sensing electronics are powered by the 2V supply for low power dissipation, and 2.7V is used to drive the switched-capacitor circuits.

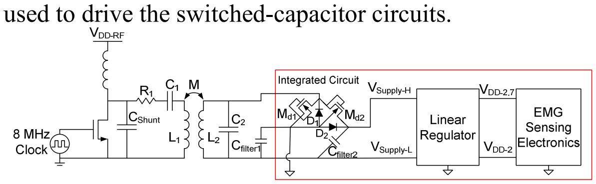

Figure 3. Simplified wireless powering system

A 3.5 µH, 6-mm-diameter implantable planar spiral coil consisting of 23 turns of 36 AWG enameled wire was built to fit around a 2 mm x 2 mm IC for a compact design. Computer simulation and bench-top characterization verify that, with a 2 µH, 10-turn spiral coil exhibiting a 2-cm outer diameter, a DC power transfer efficiency of 10% can be achieved at 8 MHz operation frequency with a 1 cm separation distance between the two coaxially aligned coils and 83 µA current loading from the regulated 2V supply [5]. The chosen frequency is also desirable for a negligible RF power dissipation in the tissue between the coils [6]. Table 1 lists the value of components used in the wireless powering system design.

Table 1. Wireless powering system parameters value

External System Parameters		**Implanted System Parameters**	
Parameter	***Value***	***Parameter***	***Value***
L_1 (10-turn spiral coil)	2 µH	L_2 (23-turn 6mm coil)	3.5 µH
C_1	408 pF	C_2 (off-chip)	100 pF
C_{Shunt}	188 pF	$C_{Filter1}$ (off-chip)	1 nF
R_1	4.2 Ω	$C_{Filter2}$ (on-chip)	100 pF
RF Powering Frequency	8 MHz		

Implantable epimysial Pt-Ir EMG sensing electrodes developed by the Cleveland FES Center are interfaced with an integrated CMOS low-noise, low-power differential amplifier based on a subthreshold design. EMG signals of interest for prosthetic control exhibit a bandwidth of approximately 1 kHz with a maximum energy contained within 250 Hz, and maximum signal amplitude of 10 mVpp. The minimum detectable EMG signal is typically 5 µVpp. The front-end amplifier is capacitively coupled to the sensing electrodes, as shown in Figure 4, with a -3 dB bandwidth of 1 kHz. A fully differential architecture achieves a high common-mode rejection ratio, critical for sensing *in vivo* real-time physiological signals. The EMG interface amplifier is designed as a telescopic operational transconductance amplifier (OTA) with a PMOS differential input pair operating in the subthreshold or weak inversion region. Subthreshold design technique [7] is used to determine the amplifier bias current of 0.85 µA to achieve an input-referred noise floor of 63 $\mathrm{nV}/\sqrt{\mathrm{Hz}}$, required for the system design specification [5].

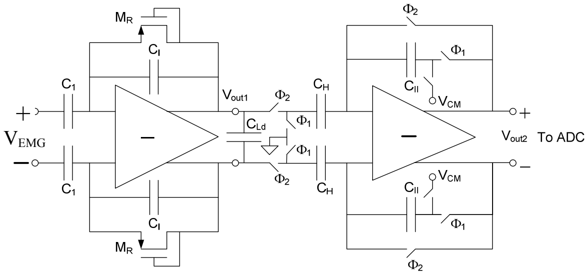

Figure 4. EMG interface electronics

DC baseline stabilization of the OTA is achieved by using active MOS resistors in the feedback path in parallel with the feedback capacitors. This sets a high-pass filter with a corner frequency of approximately 20 mHz, and provides a DC path from the amplifier output to its input. The output common-mode level of the OTA is set by a continuous time common-mode feedback circuit. The output of the EMG front-end amplifier is then sampled by a switched-capacitor 2.5x differential gain stage, which consumes 3 µA from 2 V supply, as shown in Figure 4. The 2.5x gain stage drives a fully differential 11-bit algorithmic ADC, consuming 6 µA from 2 V. This ADC topology is selected for its low power dissipation, adequate resolution for the proposed application, and low digital design complexity resulting in a small chip size in a 1.5 µm CMOS process. A ratio-independent multiply-by-two scheme implemented by sampling the input voltage twice is used to avoid capacitor mismatch limitations. The redundant signed digit (RSD) scheme employed by the ADC also improves its immunity to linearity errors caused by offset voltage and comparator accuracy. The digital output of the ADC is Manchester-coded and transmitted to an external transceiver by passive digital phase-shift keying (PSK) modulation scheme over the same wireless link used for RF powering. PSK modulation is implemented by digitally detuning the secondary (implanted) tank by means of a switching capacitor. This detuning action reflects a complex impedance back to the primary tank of the external transceiver, which can be recovered by a phase detection circuit [4, 5].

3. Measurement Results

The custom designed EMG sensing electronics are fabricated in a 1.5 µm (2-metal, 2-poly) CMOS process. Figure 5 presents a micrograph of the integrated circuit, occupying 2.2 mm x 2.2 mm area and consisting of an EMG front-end amplifier, 2.5x switched-capacitor gain stage, algorithmic ADC, RF-DC power conversion electronics, PSK telemetry circuit, and digital control block. The implantable electronics consume 83 µA in total from 2.1 V and 2.7 V supplies.

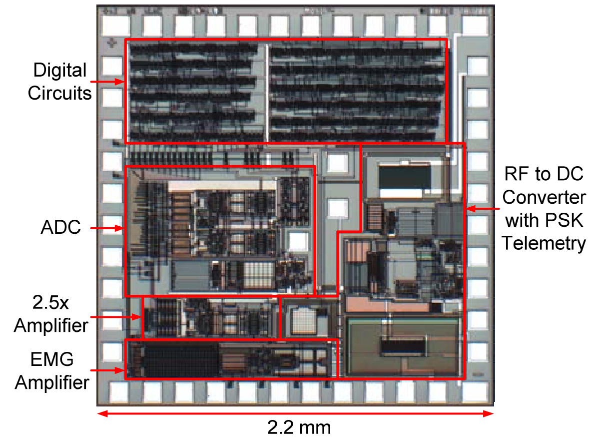

Figure 5. ASIC micrograph

Figure 6 illustrates the package arrangement of the prototype EMG sensing IC interfaced with the RF powering / data telemetry coil over a thin, flexible substrate. The packaged electronics will be encapsulated with an epoxy resin followed by silicone coating, and then interfaced with Pt-Ir epimysial EMG sensing electrodes.

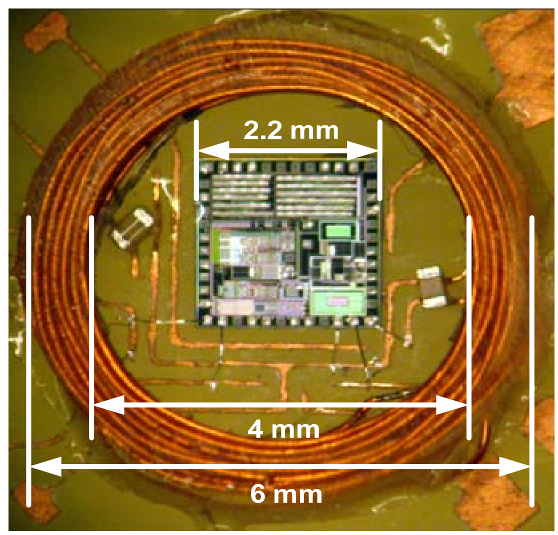

Figure 6. Prototype EMG sensing microsystem

Figure 7 presents the measured output thermal noise floor of the EMG amplifier, from which the input-referred noise can be calculated as 78 $\mathrm{nV}/\sqrt{\mathrm{Hz}}$ The amplifier exhibits a low-pass corner frequency of approximately 920 Hz with a $1/f$ noise corner frequency of 15 Hz, closely matching the design specifications. Figure 8 is the measured ADC output spectrum with a 500 $\mu V_{P\text{-}P}$ differential tone at 137 Hz applied to the EMG amplifier input. This tone represents a typical EMG signal level and frequency.

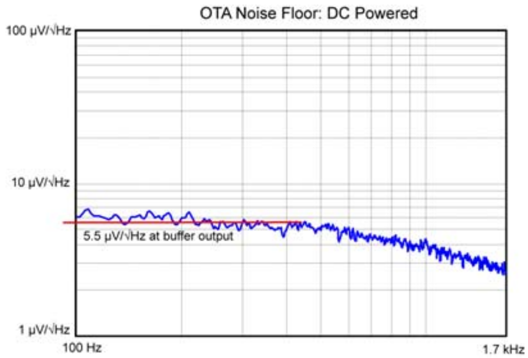

Figure 7. EMG amplifier output noise floor

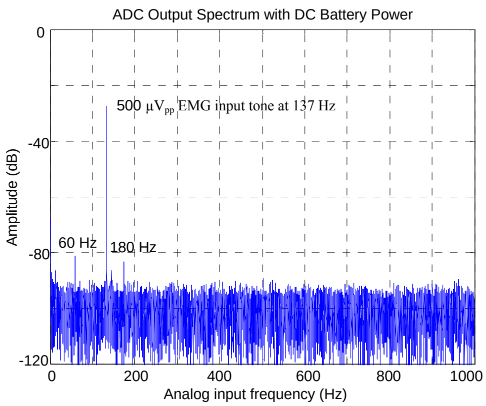

Figure 8. ADC output spectrum

The overall microsystem is charactarized for wireless powering and data telemetery as shwon in Figure 9. The implantable microsystem is powered wirelessly and transmitts Manchester-coded test data back to the external transceiver as plotted in Figure 10. The prototype systme achieves a resolution of 8.7 bits when wirelessly powered. Animal implant experiments have been successfully conducted to demonstrate EMG signals from a laboratory rat hind limb can be detected and wirelessly transmitted to an external receiver as shown in Figure 11. The baseline variation is due to measurement artifacts, which can be suppressed by post-processing and filtering.

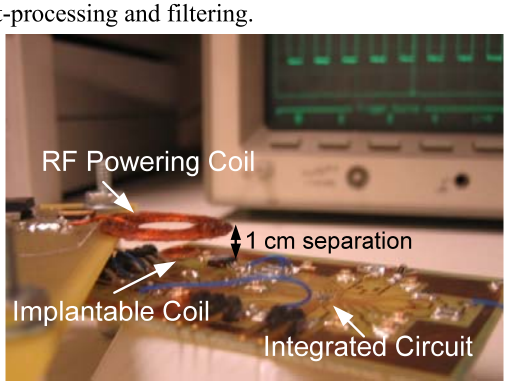

Figure 9. Microsystem wireless test setup

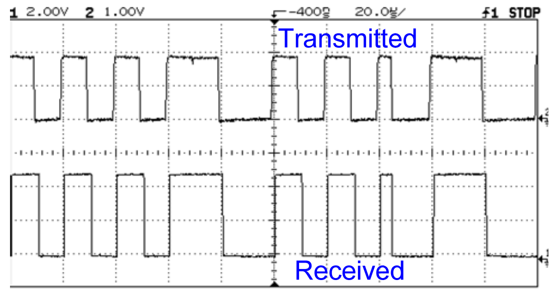

Figure 10. 50 kbps Manchester-coded PSK data telemetry

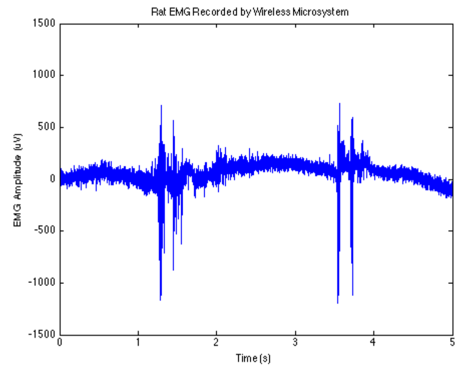

Figure 11. Wirelessly received EMG from rat hind limb

4. Conclusion

An implantable EMG sensing microsystem with wireless RF powering and PSK digital data telemetry for real-time monitoring of myoelectric signals is developed. The overall system demonstrates functionality adequate for obtaining a detailed *in vivo* EMG waveform suitable for intelligent powered prosthetics control.

References

[1] R Z. M. Nikolic, D. B. Popovic, R. B. Stein, Z. Kenwell, "Instrumentation for ENG and EMG recordings in FES systems," *IEEE Trans. Biomed.ical Engineering*, v. 41, nr. 7, pp. 703-706, 1994.

[2] N. Chaimanonart, D. J. Young, "Remote RF powering systems for wireless MEMS strain sensors," *IEEE Sensors Journal*, v. 6, nr. 2, pp. 484-489, 2006.

[3] M. Zimmerman, N. Chaimanonart, and D. J. Young, "*In vivo* RF powering for advanced biological research," *the 28th Annual International Conference of the IEEE Engineering in Medicine and Biology Society* (EMBS '06), 2006, pp. 2506-2509.

[4] N. Chaimanonart, M. Suster, and D. J. Young, "Two-channel data telemetry with remote RF powering for high-performance wireless MEMS strain sensing applications," *IEEE Sensors Conference*, October 2005, pp. 285-288.

[5] B. D. Farnsworth, R. J.Triolo, and D. J. Young, "Wireless Implantable EMG Sensing Microsystem," *IEEE Sensors Conference*, October 2008.

[6] IEEE Standard for Safety Levels with Respect to Human Exposure to RF Electromagnetic Fields IEEE/ANSI c95.1.

[7] R. R. Harrison, C. Charles, "A low-power low-noise CMOS amplifier for neural recording applications," JSSC, v. 38, nr. 6, pp. 958-965, 2004.