A 27μW subcutaneous wireless biosensing platform with optical power and data transfer

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ABSTRACT — **We present a batteryless, 27uW barely subcutaneous sensing platform using optical power and data links. We demonstrate an 8.1 mm x 3.2 mm implantable tag powered by an 850nm infrared source through the skin barrier. Measurements using pig skin indicate that optical power transfer through skin achieves a 4.9% efficiency, which is higher than mm scale inductive power links. As a proof-of-concept demo, we sense and transmit temperature to an external body worn reader using a fully optical power and data link.**

Index Terms **— Optical power transfer, optical data transfer, sub-cutaneous implant.**

I. INTRODUCTION

"Barely subcutaneous" sensors are implanted in a minimally invasive location just below the skin and provide a direct sensing interface to tissue, fluid, and electrophysiology to enable continuous healthcare monitoring [1]. Ease of sensor placement and aesthetics severely constrain the dimensions of the implant to the mm-scale and necessitate battery-free operation. Currently, most batteryless implanted systems use RF/inductive power and data transfer. While most wireless power transfer systems use a frequency of 13.56 MHz needing larger coils, recently, it has been shown that the optimal power transfer frequencies for mm sized coils is in the GHz range [2]. However, even when operating at the optimal GHz frequencies, power transfer efficiencies for mm sized coils are less than 0.1% [2][3]. In this work we show that optical power transfer can achieve better efficiency for a barely subcutaneous system and demonstrate a mm scale, batteryless, optically powered implant with bidirectional optical telemetry (see Figure 1).

A few papers in the past have proposed optical links for either power or data transfer. [4] presents an integrated bio sensor which runs off 10nW of energy harvested from ambient light but lacks a wireless transmitter. [5] Demonstrates a 16 Mbps optical link for neural recording while transferring power using an inductive link. This system uses off-the-shelf components and the systems size is of the order of cm^2. [6] demonstrates a 2Mbps optical link through the cornea from an external reader to implant while being powered by a 13.56MHz inductive link in the context of an epi-retinal prosthesis. In contrast to existing works, we present a mm-scale realization of an optically powered sensing platform that achieves bidirectional optical telemetry.

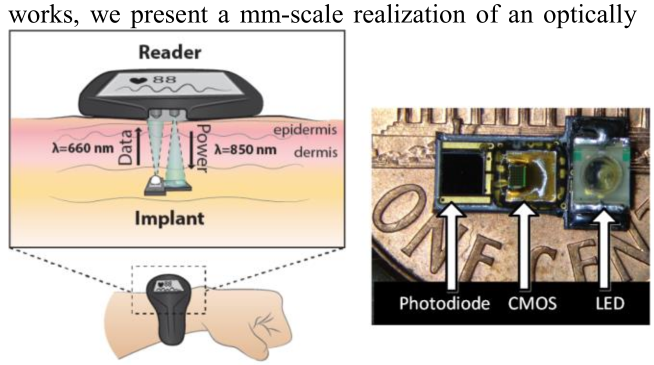

Figure 1 . Proposed subcutaneous implant

II. OPTICAL POWER TRANSFER

A barely subcutaneous implant, which is separated only by the skin barrier from the external world, opens up an alternative to inductive power transfer: optical power transfer. Human skin has a transmittance of 20% to 40% in the near-infrared wavelengths (700nm to 1000 nm) [7], which can be exploited for power and data transfer.

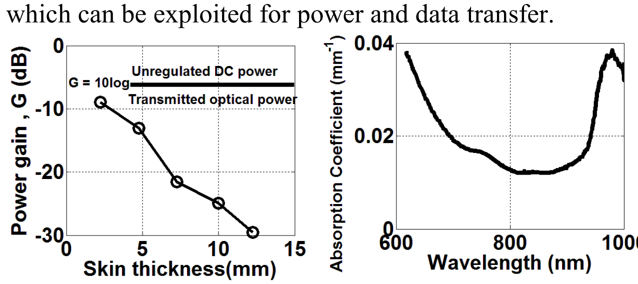

Figure 2. Measured power gain across pig skin thickness, ex-vivo human skin absorption coefficient from [7]

Measurements using pig skin indicate that we can achieve power transfer efficiency (optical power to usable DC) of ~4.9% using an 850nm IR emitter and 2mmx2mm silicon photodiode across a 4.75mm thick skin. Figure 2 shows the measured power gain (optical to unregulated DC) across multiple layers of pig skin. Though not suitable for deep implants, optical power transfer outperforms RF/inductive transfer at distances of 2-5mm which is the target reader-to-implant separation for this application.

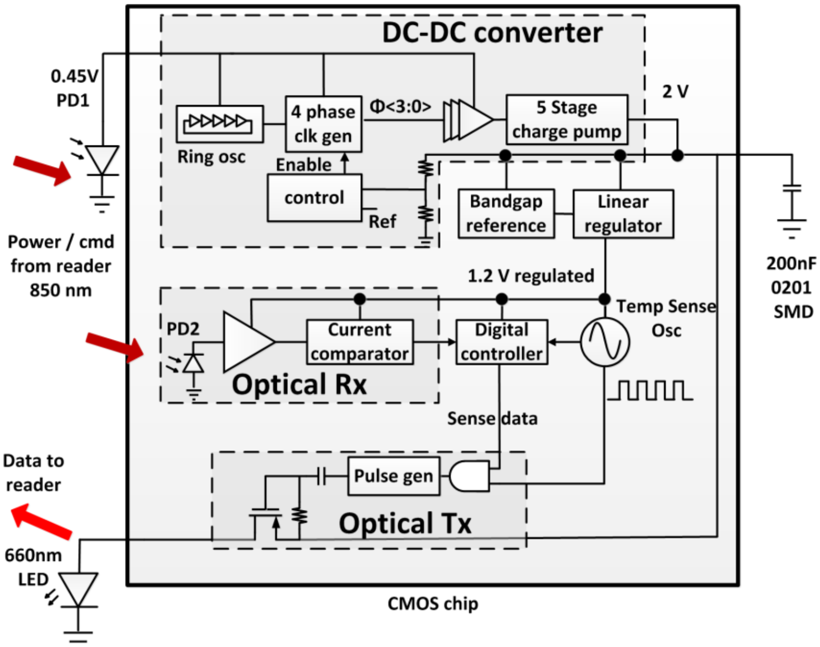

Figure 3. Architecture of the proposed sensing platform

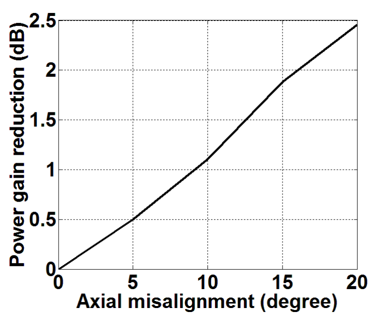

Figure 4. Effect of angular misalignment on power transfer efficiency

A key consideration for an optical power link is the misalignment between the source and the receiver. Figure 4 shows the measured efficiency loss as a function of angular misalignment. For a 10 degree tilt between the source and the receiver planes, we measured an efficiency reduction of ~1dB. While the angular misalignment is relatively benign, misalignment in X-Y position is a function of the illumination cone and reader-implant separation. Fortunately, the burden of X-Y alignment can be pushed to the reader by using an array of optical transmitters covering a wider area of skin. The reader can intelligently infer the position of the implant beneath the skin and selectively activate transmitting LEDs in the transmit array.

The American National Standards Institute (ANSI) prescribes a Maximum Permissible Exposure (MPE) of 400 mW/cm^2 to human skin at 850nm for long term exposure [8]. As a point of reference, direct sunlight has a power density of 100 mW/cm^2 with about 50% of that power in infrared wavelengths. If we assume an illumination area of (2x2) mm^2 for a miniature implant, the transmitted optical power is limited to a maximum of 16 mW. With 4.9% power transfer efficiency, up to 0.8 mW could be harvested. If we assume a conservative exposure limit of 10% of MPE, all sensing and transmission functionalities in the implant should operate within a power budget of <80uW, sufficient for many sensing modalities (glucose sensing, for example). The next section describes the architecture and the circuits enabling such a low power sensing platform.

III. IMPLANT ARCHITECTURE AND CIRCUITS

Figure 3 shows the architecture of the sensing platform. Our system runs solely from optically transferred energy from a (2x2) mm^2 silicon photodiode and achieves bidirectional optical telemetry with an external reader.
The reader transmits a modulated 850nm optical signal which provides power and data to the implant. The transmitted optical signal illuminates two photodiodes, PD1 and PD2 on the implant. PD1 is a 4 mm^2 Si photodiode (Avago technologies) which harvests the optical energy and powers the entire implant. Upon sufficient illumination, this photodiode generates around 0.45V. A switched capacitor boost converter generates a

2V DC rail. A low power bandgap reference and a linear LDO operating from the 2V supply generate a regulated 1.2V.

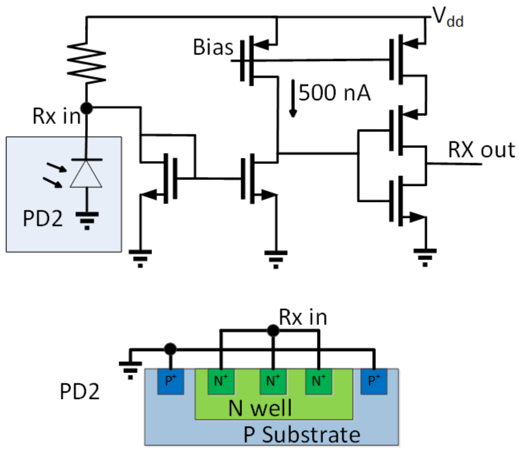

Figure 5. Fully integrated AFE and cross section of the on-chip photodiode

Optical downlink data from reader to implant is detected by PD2, an on-chip 0.7mm x 0.8 mm CMOS photodiode made from an N-well/P substrate junction. A transimpedance amplifier and slicer senses the generated signal current from PD2 and converts it to a rail to rail digital signal. Figure 5 shows the detailed schematic of the receiver connected to the on chip photodiode. The same optical signal which provides power to the chip also illuminates the data photodiode. The on-chip photodiode generates~1-2 uA signal current, enabling a slicer-based data detection. The signal from PD2 is mirrored and compared to a reference current of 500nA. Current starved CMOS inverters following the current comparator square up the signal. The receiver consumes 1.5uA from the regulated 1.2V supply. A digital block reads the output of the receiver and interprets the commands from the reader. In this chip we use a frequency counting-based command structure as a proof of concept. The digital block consumes 2.55uW.

The implant uplinks sensed data to the reader through a 16kbps 660nm optical link using an on-implant 3.5 mm x 2 mm 660 nm LED. The transmitter is powered directly from the 2.0 V supply and consumes about 5uW while transmitting with 3% duty cycle.

The proposed sensing platform can be used to measure and transmit any biological signal of interest. In this work as a proof-of-concept demo, we measure temperature and transmit it to the reader via the optical link. Temperature measurement is based on the linear frequency to temperature characteristics of a sensor oscillator. The oscillator-based temperature sensor nominally consumes 600 nW.

IV. Implementation and Measured Results

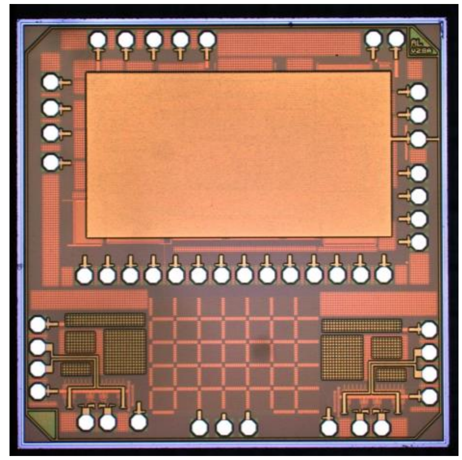

Figure 6. Die photo of the CMOS IC

All the circuits described in the previous section have been integrated in a 1.5mm x 0.9 mm CMOS IC fabricated using the IBM 130nm process (Figure 6). Detailed circuit characterization of the stand-alone CMOS IC was first done. An 8.1 mm x 3.2 mm prototype of the sensing platform, (Figure 1) housing the CMOS chip, power photodiode, PD1 and a miniature LED was then realized and tested with a companion reader board.

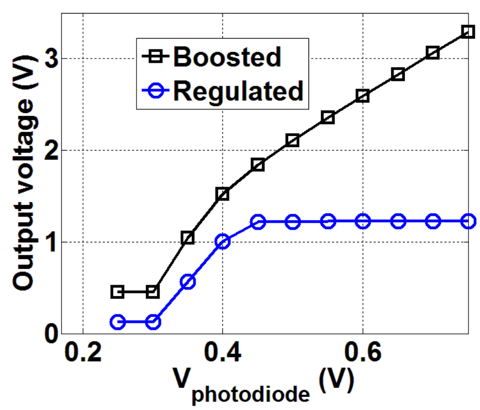

Figure 7. Measured power train performance

Figure 7 shows the measured outputs of the boost converter and the 1.2V regulator. Figure 8. shows the transmitted sensor response over temperature after two point calibration. Figure 9 shows the measured LED voltage at the optical transmitter of the implant. The reader separated from the implant by a pig skin powers up the implant and faithfully detects the optically transmitted signal from the implant (Figure 9). Table1 shows the power consumption of various blocks of the proposed system.

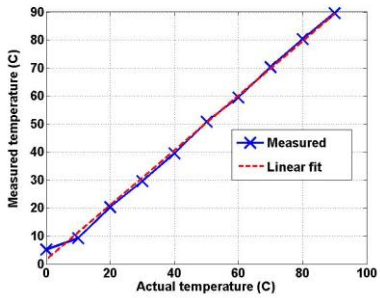

Figure 8. Measured temperature sensor response

Table 1. Power consumption of the sensing platform

Block	Power (µW)
Optical Rx	1.8
Bandgap + LDO	5
Digital	2.5
Optical Tx	5.3
Temp sensor	0.6
Boost converter	11.6
Total	**27**

Table 2. Performance comparison with prior wireless power transfer works

Performance comparison			
Performance	**ISSCC 2009[2]**	**JSSC 2013 [3]**	**This Work**
Coil/ photodiode size	2mm x2mm	0.5mm x 0.25 mm	2mm x 2mm
TX Power	250 mW	50 mW	13 mW
Medium	Bovine muscle	Air	Pig skin
Frequency / λ	915 MHz (RF)	1500 MHz RF	850nm optical
Tx – Rx distance	15 mm	1mm	4.75 mm
Available power (unregulated DC)	186 uW	10.5 uW	648 uW
Efficiency (%)	0.074	0.021	4.9

Table 2 compares optical power transfer with previous mm-scale wireless power transfer works.

The presented work to the best of the authors' knowledge is the first single-chip bio-sensing platform to demonstrate optical power transfer with full duplex optical telemetry.

ACKNOWLEDGEMENT

The authors acknowledge Prof. Yu-Te Liao and Dr. Yi Chun Shih for their valuable feedback on the design.

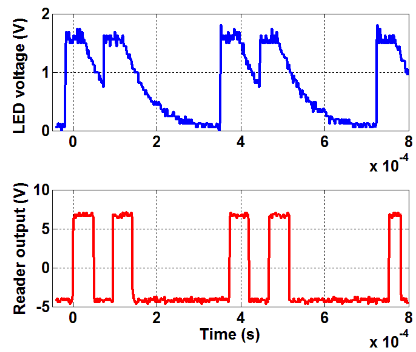

Figure 9. LED voltage at implant (Top) , Detected digital output at reader separated by pig skin from implant (bottom)

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