Design of a Miniaturized Wireless Blood Pressure Sensing Interface Using Capacitive Coupling

Ammar Aldaoud, Callum Laurenson, Francois Rivet, Mehmet R. Yuce, and Jean-Michel Redouté

Abstract—This paper presents the implementation of a miniaturized wireless blood pressure sensor interface. The system uses capacitive coupling in order to transmit the data, as well as wireless inductive powering. Designed for a bit length of 6.4 µs, the average power consumption of the device has been measured to be 20.5 µW and 2.85 mW in air and phantom material, respectively. The miniaturized sensor interface circuit consists of two plates with a diameter of 2.5 cm, which are connected by means of a thin wire; the device’s maximum thickness is 5 mm.

Index Terms—Capacitive coupling, inductive powering, wireless implantable sensor interface.

I. INTRODUCTION

According to the Australian Institute of Health and Welfare, 2.1 million Australians suffer from high blood pressure [1]: being responsible for 13% of deaths globally, high blood pressure remains the main cause for morbidity and mortality worldwide. Hypertension puts subjects at serious risk of cardiovascular and heart diseases, damages and narrows the arteries, causes aneurysms and strokes, and is even linked to dementia. Long-term analyses based on blood-pressure measurements throughout day and night as well as outside of the clinical surrounding are an important method for physicians to find anomalies in the blood pressure trend, while excluding side-effects such as the white-coat or masked hypertension, in order to determine the right diagnosis and adequate therapy for each patient [2]. However, having regular blood pressure checks throughout the day is not only impractical, but also time consuming and impairing to the subjects’ daily activities. Additionally, arterial pressure changes very quickly in response to stress, physical exertion, diet, temperature, and many other parameters. This underlines the need for an implanted continuous blood pressure monitoring system in at-risk patients.

This paper describes a proof-of-concept design for a miniaturized interface that measures blood pressure and transmits the data to an external controller. The system uses capacitive coupling in order to transmit the data: it samples the blood pressure at a frequency of 33 Hz and uses a bit length of 6.4 µs to transmit the data. The employed modulation is ON–OFF keying (OOK), in which the carrier signal is switched ON and OFF in order to represent bits 1 and 0, respectively; this technique allows a straightforward and power efficient implementation. A wireless inductive link is foreseen to power the interface circuit as well as to charge an internal supply buffer capacitor. The average power consumption of the interface has been measured to be 20.5 µW and 2.85 mW in air and phantom material, respectively.

This paper is organized as follows. The capacitive coupling which is used for the data transmission is described in Section II. The design of the miniaturized interface circuit is explained in Section III. The corresponding data receiver and the wireless power transmitter designs are presented in Sections IV and V, respectively. The measurements are reported in Section VI.

II. CAPACITIVE COUPLING

Biological tissue is a heterogeneous material where the dispersion changes with frequency [3]. Previous works have illustrated that human tissue is very good at conducting signals below 10 MHz [4]. At higher frequencies, the conductivity of the human body increases dramatically, which means that more energy is absorbed [5]. Additionally, reported in-vivo measurements have shown that the relative permittivity of human skin varies between 1000 at 1 MHz and down to 250 when the frequency increases to 10 MHz [6].

Inductive coupling is quite common in biomedical implants, but it presents two primary disadvantages. First, an inductor creates a magnetic field in all directions, which can potentially interfere with other electronics on an implant [7]. Second, constructing a planar or flat inductor with a suitably high quality factor is challenging and is especially difficult, given the small size of the implant. A possible alternative is to use capacitive coupling. This involves two implanted plates which could couple with two other plates on an external receiving device, where the human tissue acts as a conductive dielectric of which the permittivity is highly dependent on the frequency [see Fig. 1(a)]. The resulting equivalent electrical model is depicted in Fig. 1(b). In the latter figure, the following quantities are defined.

1) \( C_c \) is the coupling capacitance between a transmitting and receiving plate.
2) \( C_p \) is the parasitic capacitance between the plates on the implant and the plates on the receiver.
3) \( R_p \) is the resistance between the transmitting and receiving plates.
4) \( R_p \) is the resistance between the plates on the implant and the plates on the receiver.

This electrical model assumes a uniform dielectric material (“human tissue”), meaning that the dielectric impedance between each implant plate and its respective receiving plate is the same: it also assumes that the parasitic capacitances between the plates on the implant and the plates on the receiver are identical. The resulting transfer function is equal to

\[
\frac{V_{RX}}{V_{TX}} = \frac{\frac{R_p}{1+j\omega R_p C_p}}{1+j\omega R_p C_p + \frac{2R_p}{1+j\omega R_p C_p}} \tag{1}
\]
where $V_{TX}$ and $V_{RX}$ is measured on the transmitter and receiver side, respectively. It can be observed that the parasitic impedance between the signal and ground planes should ideally be as small as possible in order to maximize the coupling as well as the corresponding output signal. This is achieved by increasing the distance between the capacitive plates on the implant and on the receiver; however, the greater the distance between the plates, the larger and the less practical the implant becomes. For this implementation, a distance of 10 cm between the signal and common plates was chosen as a compromise between these two conflicting requirements.

III. MINIATURIZED BLOOD PRESSURE SENSOR INTERFACE DESIGN

A. Introduction

The miniaturized blood pressure sensor interface requires three essential criteria to be filled. First, the circuit implementation should be made as small as possible to demonstrate that it could potentially be implanted; second, it should consume as little power as possible as well as be wirelessly powered so as to be fully implantable without the need for a battery or external connections; third, it should transmit the measured data wirelessly to an external controller at a frequency of 33 Hz, which corresponds to the sampling frequency reported by state-of-the-art work [8].

The top-level schematic of the miniaturized sensor interface is depicted in Fig. 2. The circuit consists of a sensor, a differential amplifier, a microcontroller, and a wireless power unit. As two capacitive plates are required, the device is constructed on the flip side of the first plate, while the active signal is transmitted to the second plate (refer to Fig. 1). The interface is powered wirelessly and transmits data using capacitive coupling, as explained in Section II. The circuit samples and transmits the pressure at 33 Hz.

A picture of the device without the sensor and supercapacitor is shown in Fig. 3: observe that its size is comparable to an Australian 10 cent coin. Including the supercapacitor of 47 mF, as well as the coil, results in a thickness of 5 mm (see Fig. 4).

B. Pressure Sensor

In the final implementation, a miniaturized microelectromechanical system (MEMS) sensor will be used to measure the blood pressure. In essence, all MEMS pressure sensors are based on thin diaphragms that deflect due to a pressure difference: the
sensing of these sensors is capacitive, either piezoresistive [9]. In the present system, a piezoresistive pressure sensor was chosen and the interface circuit was designed accordingly.

Selecting a sensor that measures the pressure of a fluid results in a proof of concept that reflects the real scenario more accurately: a fluid also has a higher relative permittivity compared to air, which complements the capacitive coupling between the transmitter and receiver. However, the full device would have to be sealed, which, in turn, would make it inaccessible for electrical measurements and debugging. In order to develop a proof-of-concept design, air pressure was eventually chosen as it does not require an insulated and closed-off sensor interface. In order to save power, an analog pressure sensor was selected (NPP-301-200AT from GE): this sensor has a pressure sensing range between 0 and 30 psi. Blood pressure does not typically go above 250 mmHg or 5 psi, meaning this sensor has a reasonable range for the target application.

C. Differential Amplifier

The sensor is piezoresistive and records the pressure using a Wheatstone bridge configuration, meaning that a differential amplifier is required in order to amplify the weak sensor signal. Since the bandwidth of the entire system is limited, a low-power operational amplifier can be used. The LT6003 (Linear Technology) opamp comes in a single package and does not draw more than 1 μA: the gain bandwidth product is equal to 2 kHz.

D. Microcontroller

The microcontroller chosen for this design was the MSP430G2231 mixed-signal microcontroller from Texas Instruments. It has two internal clock sources, namely the digitally controlled oscillator (DCO) and the very low power oscillator. The former can be configured to run at frequencies between 0 and 20 MHz, with approximately 100 discrete steps between these frequencies: the latter is nonconfigurable and outputs a frequency 12 kHz with very poor accuracy. Since a slow clock consumes far less power, it can be utilized when the microcontroller is placed into one of its many low power modes. By turning ON parts of the microcontroller only when required and using slower clock speeds when possible to save power, the consumed power has been drastically reduced.

In this design, the MSP430G2231 microcontroller’s clock is used as the carrier for a signal, which simplifies the overall architecture of the interface, thereby saving power and area. The submaster clock (SMCLK) which is derived from the DCO can be connected directly to an output pin, thereby making it possible to transmit data using OOK by switching the pin between the SMCLK clock and high impedance. An initial setup was made using a 3-V power source and the MSP430G2231 with pin 6 connected to the SMCLK oscillating at 8 MHz. When pin 3 of the microcontroller is high, the amplifier and sensor are turned ON. The ADC takes a reading via pin 4 and resets pin 3 to disable the amplifier and sensor to save power.

These results were promising and provided a proof of concept. However, it seemed necessary to make sure that the coupling was not due to ground connections between the oscilloscope and the signal generator. To test this, the MSP430G2231 was programmed to output a 8 MHz on pin 6 and a 1-cm² conductive plate was attached to this pin. Running the microcontroller off batteries meant that the ground connection between the signal generator and the oscilloscope could potentially be ruled out as a source of coupling. The receiving plate was a 10-cm² blank PCB. Measurements confirmed that the received voltage is proportional to the distance between both plates, since the latter behave as a parallel plate capacitor [10]. Measurements in the phantom material confirmed that the receiver would be required to amplify signals with an amplitude of not less than 40 mV.

E. Wireless Power Receiver

Space is at a premium, meaning that the wireless power receiver needs to take up as little space as possible. A 1-MHz carrier frequency was chosen as it is high enough to ensure that the resonator can be small, while being straightforward to generate on the transmitter side. Five turns of 0.8-mm enamel-coated copper wire were wound around a Teflon bobbin, specifically made to house the inductor and sensor interface PCB. The inductance was measured to be 2.8 μH. A Schottky diode was used to rectify the received sinusoid, and the Zener diode ensures the voltage does not exceed 3.3 V. A 47-mF supercapacitor was used to store the required energy so as to provide the device with a degree of autonomy.

IV. IMPLEMENTATION OF THE DATA RECEIVER

The receiver is required to pick up signals with a magnitude as low as 40 mV while filtering out common-mode noise sources, such as the 50-Hz power line interference. The receiver achieves this with four stages. The first stage consists of an LC tank that attenuates 50-Hz noise significantly. The next stage is an active first-order bandpass filter with a center frequency of 8 MHz and a gain of 27 dB. This selects and amplifies the desired signal and removes more unwanted noise. For an input signal of 40 mV, this stage outputs a signal with a magnitude of not more than 900 mV. In the third stage, the signal is inverted and amplified with a gain of 33 dB, which causes the signal to reach and clip at the 3-V supply rails: since nonreturn to zero encoding is being employed, this is desirable. In the fourth stage, the OOK signal is demodulated by a diode detector. The data receiver front-end is depicted schematically in Fig. 4. The complete receiver has an LMV761 comparator (Texas Instruments) connected to the output, which ensures that the signal is as noise free as
Fig. 6. Implementation of the data receiver.

Fig. 7. Top view of the complete setup.

possible (not shown in Fig. 5). This binary signal is then fed into a PIC16F628A 8-bit microcontroller (microchip), which outputs the data to an LCD screen as well as to a serial to USB interface, which allows the recorded data to be stored on a PC. The construction and implementation of the receiver are shown in Fig. 6.

V. WIRELESS POWER TRANSMITTER

On the transmitter side, a parallel LC tank is driven by dual H-bridge L293 drivers (Texas Instruments). The latter are driven by two TTL logic-level square waves, each 180° out of phase with the other. The transmitter coil was made using transformer wire, as it has extremely low equivalent series resistance due to its thickness, giving the transmitting inductor a high quality factor. Two turns were wound around a PVC pipe with a diameter of 200 mm. The transmitting inductor has a measured inductance of 1.5 μH.

VI. MEASUREMENTS

The full setup is depicted in Fig. 7. The data receiver with LCD screen displaying the sensed pressure is at the top left side. The data receiver requires two metal plates; the second metal plate is underneath the LCD screen and receiver circuit. The powering inductor is shielded with black tape and located at the top center. The miniaturized sensor interface is placed in a rectangular glass box, which rests on the powering inductor. The phantom material is kneaded in the glass supporting box (not shown here). The H-bridge driving circuitry for powering the inductor is situated in the top right corner. At the bottom right is the diaphragm pump, with the plastic tube disconnected for the photo and rolled up in the pump box. This pump pressurizes the air in the tube, which fits over the pressure sensor vent embedded on the interface PCB (center). At the bottom center, a strapped Texas Instruments launchpad is connected to a project box (bottom left) housing all external connections to the system (i.e., complementary square wave used to drive the H-bridge for the wireless power transmitter, the power supply to the H-bridge and diaphragm pump and the pump switch) as well as programming pins used for (re)programming the microcontroller on the interface circuit.

In a first measurement, the receiving coil was wirelessly powered through air: the distance between the transmitting coil and the receiving coil on the miniaturized interface was measured to be 8 cm. The link efficiency was found by taking the quotient of the input and output power

\[ \eta = \frac{P_{RX}}{P_{TX}} = 0.25\% . \]  

(2)

The power link is clearly weakly coupled [11]. However, when the receiving coil was mounted on the interface PCB, and embedded in the phantom material, the measured power link efficiency decreased to 0.01%. This can be attributed to the ground plane in the center of the inductor that is used for capacitive data transfer: this plane attenuates the magnetic field, causing the efficiency of the inductive power link to drop. One solution could have been to keep the ground plate and the inductor separated by some distance, but this would increase the size of the device. The ground plate could also potentially be reduced in size to allow more magnetic field to couple with the inductor. Either way, there is a tradeoff to be made between the plate sizes and inductive link efficiency.

An approximate phantom material was constructed using play dough. The properties of this phantom material were derived from using a plate capacitor which was constructed out of two copper plates, each 80 mm², separated by 2.25 mm, and using air and the phantom material, respectively, as the dielectric medium. The quotient of the latter values gave the magnitude of the complex permittivity (\( \epsilon_r(\omega) \)) [12]: at 8 MHz, it was found to be equal to

\[ \epsilon_r(\omega) = \epsilon'_r(\omega) - j \epsilon''_r(\omega) = 5.05 - j \cdot 20.38 \]  

(3)

where \( \epsilon'_r(\omega) \) relates to the stored energy within the medium, while \( \epsilon''_r(\omega) \) reflects dissipation of energy within the medium due to conduction and polarization losses. Using the same setup, the equivalent series resistance of the resulting phantom material capacitor was measured to be 30 Ω. As can be seen, the designed phantom material behaves much worse compared to human tissue: the so-formed dielectric is highly conductive, which exacerbates the power losses, while the relative...
permittivity is considerably lower than the one presented by human tissue (refer to Section II).

The data transmission determines the average consumed power: for each sampled measurement, there is one start bit (high), 9 data bits, as well as one stop bit (low). The start bit and the high data bits consume power, while the low data bits and stop bit draw a negligible current. In order to make a fair comparison, it was decided to transmit the start bit as well as four high-data bits when taking the measurements, as this figure represents the average transmitted bit code. The bit length was programmed to be equal to 6.4 $\mu$s. Using a bit length of 6.4 $\mu$s and transmitting 5 high bits per sample at 33 Hz, the average power consumption of the device was measured to be equal to 20.5 $\mu$W and 2.85 mW in air and phantom material, respectively. This difference is explained by considering that the phantom material increases the parasitic coupling between both plates on the receiver and transmitter side since the phantom material, as human tissue, has a finite resistance: the latter, in turn, increases the power consumption, confirming the observations made in Section II. The average power consumption corresponds to a transmitted energy per “high” bit equal to 131 pJ/bit and 18.2 nJ/bit in air and in the phantom material, respectively.

Finally, although the transmitted signal is attenuated by the phantom material, the data transmission from the interface circuit to the receiver behaves as expected, as illustrated in Fig. 8. Table I presents a state-of-the-art review situating this study.

### VII. Conclusion

This paper has presented the implementation of a miniaturized wireless blood pressure sensor interface. The system uses capacitive coupling in order to transmit the data, as well as wireless inductive powering. Measurements validated the design simulations and calculations, and showed that, using a bit length of 6.4 $\mu$s, the power consumption of the miniaturized interface circuit has been measured to be 20.5 $\mu$W and 2.85 mW in air and phantom material, respectively.

### ACKNOWLEDGMENT

The authors would like to thank I. Reynold, M. Linzer, and A. Linzner from the Department of Electrical and Computer Systems Engineering, Monash University, Melbourne, Australia, for their help in manufacturing the prototype and setup.

### TABLE I 
STATE-OF-THE-ART REVIEW

<table>
<thead>
<tr>
<th>Reference</th>
<th>Power consumption</th>
<th>Data rate</th>
<th>Frequency</th>
</tr>
</thead>
<tbody>
<tr>
<td>[13]</td>
<td>300 $\mu$W</td>
<td>48 kb/s</td>
<td>4 MHz</td>
</tr>
<tr>
<td>[14]</td>
<td>0.2 mW</td>
<td>2 Mb/s</td>
<td>10 KHz–100 MHz</td>
</tr>
<tr>
<td>[15]</td>
<td>400 mW</td>
<td>2.4 kb/s</td>
<td>330 KHz</td>
</tr>
<tr>
<td>[16]</td>
<td>5 mW</td>
<td>5 Mb/s</td>
<td>0.3–15 MHz</td>
</tr>
<tr>
<td>[17]</td>
<td>650 mW</td>
<td>12 Mb/s</td>
<td>–</td>
</tr>
<tr>
<td>This work</td>
<td>20.5 $\mu$W (air), 2.85 mW (phantom material)</td>
<td>363 b/s</td>
<td>8 MHz</td>
</tr>
</tbody>
</table>

### REFERENCES