Abstract—Recent advances in low power integrated circuits is opening up a whole new realm of possibilities in the field of medical implants. One such possibility is to have implants which require neither wires nor batteries. Such an implant could be put in place and monitored using external equipment for the life of the patient, and it would never require later surgical operations to replace the batteries. The implant could be powered by the RF field generated by an external reader, and it would transmit its data back to the external reader using that same RF field. This paper will discuss the development of a medical implant which is designed to measure the blood pressure level profile from inside an abdominal aortic aneurysm sac. This implant is designed to collect pressure sample readings from an array of pressure sensors, and to transmit those pressure sample values to an external reading device.

Index Terms—medical implant, rectenna, RFID, wireless power transmission, micromachined pressure sensor

I. INTRODUCTION

This project explores the development of a batteryless and wireless blood pressure sensor for monitoring aneurysm intrasac blood pressure levels. This pressure sensing medical implant uses RF power energy harvesting and wireless data transmission, allowing for non-invasive monitoring of the blood pressure levels from inside the aneurysm sac (Fig. 1).

The blood pressure sensing implant uses an array of piezoresistive micromachined pressure sensors spaced evenly around the stent graft to collect data from within the aneurysm sac. The collected data is transmitted wirelessly to an external, hand held, RFID reader allowing user software to generate real time images of the pressure profile from inside the aneurysm sac. The entire aneurysm sac pressure monitoring system consists of the pressure sensing implant and the hand-held reader.

II. PROJECT MOTIVATION

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A. Implant Energy Harvesting

The implant antenna is looped around the stent in a way that maximizes the amount of area enclosed by its contour and it is tuned to resonate at the reader’s 13.56 MHz carrier frequency. The antenna collects the transmitted power from the reader antenna, and the received power, which is in the form of an oscillating current and voltage, is fed into the DC rectifier. The rectified voltage is then regulated down to 1.8 VDC to power the microcontroller and the pressure sensors.

Fig. 1. Diagram of aneurysm, implant, reader, and PC interface
B. Ultra-Low Power Microcontroller

The implant’s microcontroller is from Texas Instruments’ family of MSP430 ultra low power microcontrollers. It has an integrated 16 bit $\Delta\Sigma$ differential analog to digital converter with an 8 channel multiplexer which, in addition to being able to do conversions on the 4 external ADC channels, can also be used to measure supply voltage, temperature, and internal ADC offset error. The MSP430 also has an integrated PWM controller which is used to modulate the 13.56 MHz carrier frequency with a 250 kHz subcarrier which carries the 50 kbps digital data back to the reader. The microcontroller and the associated energy harvesting circuitry is centrally located on the implant, and everything is sealed in a rigid epoxy. The epoxy hermetically seals the circuitry while also giving necessary structural rigidity to protect the solder bonds from fracturing when the implant is collapsed and expanded during implantation into the aneurysm.

C. Micromachined Pressure Sensors

The pressure sensors are machined from pure silicon crystal, and they are cubic in shape with each side having a length of 0.65 mm. The sensors are mounted onto the PCB so that when the implant is mounted onto the stent each sensor will be spaced evenly around the circumference of the stent about midway up the length of the stent (Fig. 3). Each sensor is wire bonded to the PCB substrate, and the sensors are coated with a layer of silicone sealant (Fig. 5). The silicone sealant serves the dual purpose of protecting the bond wires while also enclosing the pressure sensor in a biocompatible enclosure. Also, because the silicone remains pliable after curing, the pressure sensor’s ability to measure changes in the aneurysm pressure is not too greatly diminished. The pressure sensors used for the implant have a pressure sensitivity range of 0-4 atmospheres of pressure [6], and the response of the sensors to pressure is a linear analog function which is only limited by the ADC’s digitizing resolution.

D. Flexible Polyimide PCB Substrate

The PCB substrate used for the implant is a flexible polyimide film substrate. Polyimide film has already been shown to possess excellent biocompatibility [7], and it is flexible enough to be collapsed down to a size which allows for the system to be implanted along with the endovascular stent into the aneurysm (Fig. 3). The implant’s shape is such that it can be sewn using surgical suture thread onto the metal filament spring support structure of the endovascular stent. The antenna which is used for energy collection and data transmission is integrated into the flexible PCB design.

Fig. 2. Pressure sensing implant functional diagram

Fig. 3. Image of an endovascular stent and the developed pressure sensing implant displaying the implant’s PCB shape which allows for mounting onto the stent.

Fig. 4. Image of the microcontroller and associated energy harvesting circuitry encapsulated in rigid epoxy.

Fig. 5. Image of the silicon micromachined pressure sensor mounted onto the flexible PCB with the point of a pencil to show scale.

IV. POWER TRANSMISSION AND DATA TELEMETRY

The implant is designed to operate without batteries, and without a wired connection to external equipment. The implant has to collect all of its operating power from the external reading device, and it has to transmit all of its data wirelessly to the reader device. Because everything about the system requires efficient coupling between the external reader and the implant, antenna design is one of the most critical components of a batteryless and wireless implant. In designing a proper antenna for a power and data transmission system, certain compromises have to be made between read range and power, and between efficiency and data rate.

\[
B = \frac{\mu_0 I}{4\pi} \int dl \times \hat{r} \quad (1)
\]

\[
dl \times \hat{r} = dl \sin \theta
\]

\[
dl = Rd\phi
\]

\[
\sin \theta = \frac{R}{r}
\]

\[
r = \sqrt{R^2 + z^2}
\]

\[
2\pi = \int_0^{2\pi} d\phi
\]

\[
B(z) = \frac{\mu_0 I}{2} \frac{R^2}{(R^2 + z^2)^{3/2}} \quad (2)
\]
A. RF Power vs. Distance

Unlike most antennas which are designed to operate in the far field several wavelengths in distance away from the antenna, this system uses near field coupling (NFC) where the power does not actually radiate away from the antenna. Instead, the energy from the transmitting antenna generates a magnetic field surrounding the antenna, and the amount of power that the implant can collect is dependent on the magnitude of this generated magnetic field. This is where the power vs. distance relationship comes into play. When the Biot-Savart [8] law is solved for wire loop of radius $R$ vs. distance from the loop $z$, a formula for the magnetic field intensity along the center axis away from the antenna can be derived (Eqs. 1 and 2). Figure 6 is a plot of (2) with the constant $\frac{\mu_0}{2}$ set equal to unity, and it shows how a smaller antenna will generate a stronger magnetic field but will not project the field very far, and how a larger antenna will generate a weaker field but will project the field further away from the antenna.

B. Power Transmission Efficiency vs. Data Rate

The system uses a reader which is transmitting a carrier frequency $\omega_0$ to generate a magnetic field, and the implant collects its operating energy from that magnetic field. The implant also modulates the amount of energy it collects from the field, this is called load modulation, and this modulates the carrier at $\omega_0$ with a subcarrier frequency at $\omega_c$ to transmit data back to the reader. By increasing the amount of energy stored in the magnetic field at resonance, the power transmission efficiency is increased. And, by increasing $\omega_c$, the amount of data which can be modulated onto $\omega_0$ is also increased. The term quality factor (Q factor) is used to define these ratios. The Q factor describes the amount of energy stored in the reactive field to the amount power dissipated to resistive losses (Eq. 3), and it also describes the ratio of the center frequency to the subcarrier bandwidth (Eq. 4). For this case of load modulation, the total bandwidth is $2\omega_c$ because the implant is transmitting with double sideband data modulation.

\[
Q = \frac{\text{Reactive Energy}}{\text{Dissipative Energy}} \quad (3)
\]

\[
Q = \frac{\omega_0}{\omega_{hi} - \omega_{lo}} = \frac{\omega_0}{2\omega_c} \quad (4)
\]

\[
\frac{\omega_0}{2\omega_c} = \frac{\text{Reactive Energy}}{\text{Dissipative Energy}} \quad (5)
\]

Equation (5) shows the trade off between data bandwidth and power transmission efficiency. For a fixed $\omega_0$, Eq. (5) shows that to increase the bandwidth $\omega_c$, the ratio between reactive and resistive energy of the system must decrease, and vice versa. Figure 7 is a plot of collected power vs. distance for the same antenna which has been set equal to three different quality factors. Figure 7 shows how an antenna which is tuned to a Q factor of 200 is able to transmit to the receiver roughly 12 dBm more (which is 16 times the amount of power) than when the antenna is tuned for Q=7. With $\omega_0 = 13.56 \text{ MHz}$, a Q factor of 200 is suitable for a bandwidth of about 33 kHz, and a Q of 7 is suitable for a bandwidth of about 1 MHz.

Fig. 6. Plot of Eq. (2) where $\frac{\mu_0}{2}$ has been set equal to 1.

Fig. 7. Plot of the collected power vs. distance. Transmitting antenna dimensions are 28cm x 35cm, and the receiving antenna dimensions are 1.8cm x 7.5cm.

Fig. 8. Photo of the implant prototype demonstrator submerged in water and transmitting a digital waveform to the external reader device for display on a laptop computer screen.

V. Conclusion

LABORATORY tests of the implant prototype demonstrate that this is a viable method for collecting data from inside the body using wireless power and data transmission. Figure 8 is an image of the implant prototype in operation showing the device’s ability to collect its operating energy from the RF energy transmitted by the external reader device, and to transmit digital data back to the reader by load modulation of the RF carrier. In figure 8 the implant prototype has been sealed in a plastic bag and immersed in a container of saline water. The implant is located roughly 15 cm away from the antenna which is transmitting 500 mW of power. The implant prototype is collecting its required energy, which at this sample rate of 1,280 sps is about 2 mW, and it is transmitting its data to the reader at 50 kbps on a 250 kHz subcarrier using on-off shift keying.
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REFERENCES


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