MICROMACHINED ENDOVASCULARLY-IMPLANTABLE WIRELESS ANEURYSM PRESSURE SENSORS: FROM CONCEPT TO CLINIC

Mark G. Allen
Georgia Institute of Technology
Schools of Electrical and Computer Engineering and Chemical and Biomolecular Engineering
Atlanta, GA 30332-0250 USA

ABSTRACT

Passive wireless pressure sensors, originally developed for use in harsh environments, have been adapted for use in the human body. The application of these sensors is as monitors of endovascularly-repaired abdominal aortic aneurysms. For this application, the devices must be permanently implanted deep within the body and be functional for the remainder of the patient's life. Microelectromechanical systems (MEMS) manufacturing technologies have been utilized to fabricate sensors with sizes and form factors suitable for endovascular delivery and permanent implantation. The sensors are interrogated with an external measurement antenna and a real-time waveform of the pressure environment is extracted. This paper reports the development and clinical demonstration of these sensors.

Keywords: pressure sensor, wireless, implantable, endovascular, aneurysm

INTRODUCTION

The use of wireless sensing to monitor physiological parameters within the body is gaining widespread interest, due in part not only to advances in medicine (which help identify promising application opportunities), but also to advances in microelectronics and micromanufacturing technology (which have made such devices practical). Physiological parameters of interest include temperature, pressure, and chemical species, among others.

Sensors implantable within the human body for the measurement and wireless transmission of physiological parameters have been discussed for almost fifty years. In 1957, the "endo-radiosonde" was reported by Mackay and Jacobson [1]. Several years later, a device for measurement of gastrointestinal (GI) pressure was described in [2]. In this device, a predecessor of today's 'smart pills', a capsule containing a pressure sensor was swallowed and used to monitor and wirelessly transmit pressure as it moved through the GI tract. A landmark effort in miniaturization of these devices, although still achieved using traditional 'macro' fabrication approaches, was discussed by Collins in 1967 [3]. In this work, a miniature 'transensor' was described for implantation in the eye for the continuous measurement of glaucoma. This device consisted of a passive LC resonant circuit, the resonant frequency of which depended on the embedding pressure surrounding the device. Such an approach allows for a large degree of compactness and greatly reduced sensor complexity, both of which are desirable for ultrareliable operation of implanted sensors. Magnetic coupling of the device to an external loop allowed the determination of the resonant frequency of the LC sensor, and therefore with suitable calibration the embedding pressure. The sensor consisted of a glass tube with flexible polyethylene terephthalate (Mylar) diaphragms bounding each side of the tube. Supported on the two diaphragms were two hand-wound spiral inductors, with a bridging interconnection between the inductors. The inductors together with their associated distributed capacitance formed the resonant circuit. A change in the embedding pressure would change the relative separation of the coils, and therefore the resonant frequency of the circuit. Devices ranging from 6 to 2 mm in diameter were successfully fabricated. Typical distances over which the resonant frequency could be determined were approximately 3 cm for the largest devices. Implantation of these devices in the eyes of approximately 70 rabbits, along with measurement of associated intraocular pressures, was successfully demonstrated.

Since this work was performed, advances in both sensor fabrication technology as well as external readout electronics have inspired a number of investigators to apply MEMS solutions to the implantable wireless sensor problem. An incomplete review of some of these efforts is given below. In 1992, Rosengren et al. [4] discussed a MEMS implementation of Collins' intraocular sensor, utilizing two anisotropically-etched and fusion-bonded wafers with an oxide interface to form a variable capacitor. This device was connected to an external coil, and ultimately coil and capacitor were embedded in silicone [5] to form a sensor approximately 5mm in diameter. This device was implanted in rabbit eyes, and successfully measured pressure with readout distances of a few millimeters. In 1995, Carr et al. described a remotely-powered pressure sensor with coils, capacitor, and a single transistor for measurement of pressure. An external induction coil powered a Hartley oscillator with a variable pressure-sensitive capacitor in the tank circuit [6]. An alternative approach described by Suster et al. utilized a tunnel diode together with a 110 mV power source as a gain element to induce self-oscillation in an otherwise passive LC resonator [7]. Passive resonant pressure sensors made from MEMS-based ceramic fabrication approaches, with alumina diaphragms and

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fired metallic inks, have been utilized in high-temperature environments [8-9]. Anodically-bonded MEMS-based passive LC resonators for the sensing of pressure have been described by Park et al. [10] and Akar et al. [11] with mm-scale overall dimensions. With the incorporation of a polymer layer, passive LC humidity sensors [12] have also been described. Humidity sensors for monitoring the health of hermetic packages in-vivo are described by Harpster [13]; ferrite was incorporated into the sensor for improved coupling (typical coupling distances 2cm) and the device was sealed within a glass/silicon housing (approximately 7x1.2x1.5 mm) for hermeticity and implanted in guinea pigs in a variety of locations. Good package stability over month time scales was observed. Passive LC pressure sensors as components of a cerebrospinal fluid shunt system were described by Yoon et al. [14].

Although most of these devices have been demonstrated on the bench, and some in animals, substantial further testing and modification is required prior to successful clinical application in humans. In particular, establishment of stability as well as increases in the distance over which the devices can be detected is required. In the remainder of this paper, an application for the use of these devices is defined, appropriate LC resonators as well as catheter-based techniques for delivery of the resonators are designed and fabricated, and clinical evidence of the utility of the devices is collected.

ABDOMINAL AORTIC ANEURYSMS

As an example of the utility of wireless pressure sensing, consider the treatment of *abdominal aortic aneurysms* (AAAs). AAAs are a weakening of the aortic wall near the junction of the aorta and iliac arteries in the abdomen (Figure 1). Rupture of AAAs is currently the thirteenth leading cause of death, and the third leading cause of death among men over 60. Its prevalence is approximately 1.5 million cases, and approximately 200,000 new cases are diagnosed annually. Undiagnosed AAAs are such a problem that in the United States, the government is considering the addition of a screening benefit for AAAs to Medicare to better identify at-risk patients.

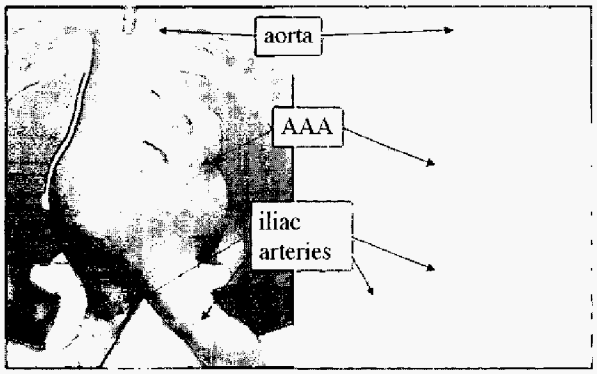


Figure 1. *Abdominal aortic aneurysm (AAA). Left: Schematic drawing showing location of AAA at the junction of aorta and iliac arteries. Right: angiogram showing presence of AAA. Darkened regions show the presence of radiopaque dye within the arterial system. To give an idea of scale, aortic aneurysms are typically treated when they exceed 5cm in diameter.*

Repair of AAAs has typically taken place in a surgical fashion, in which the weakened section of the aorta is replaced with a surgical graft. However, this approach is often traumatic for the patient, with high morbidity and mortality rates, due to anatomical constraints. At this point in the body, the aorta is located near the spine. Surgical access therefore requires a relatively traumatic procedure. A large incision is made in the abdomen, and the contents of the abdomen are displaced to provide access to the aneurysm. The surgical graft is sewn in, the abdominal contents are replaced and repositioned, and the abdominal incision is sewn up. The traumatic nature of this surgery often yields lengthy hospital stays, months before a full recovery is made, and a combined morbidity rate (early and late) of almost 15% [15].

More recently, endovascular therapy, in which catheter-based approaches are used to repair blood vessels from within, has been utilized for AAA repair. A more familiar analogous treatment is the repair of coronary vessels by use of endovascularly-delivered stents rather than open-heart surgery. In the case of AAA repair, a *stent-graft* is utilized rather than a stent (Figure 2, left). A stent-graft can be thought of as a stent (typically a metal mesh) that has had the graft material sewn onto its surface, forming a new lumen for blood flow and excluding the aneurysm wall from systemic pressure. Endovascular stent-graft repair is associated with greatly shortened hospital stays and a combined morbidity rate of approximately half that of the surgical repair [15].

Although this approach is gaining wider acceptance, the stent-grafts can suffer from endoleaks due to a variety of causes. Should an endoleak occur, the aneurysm can become repressurized, with potentially catastrophic consequences. Due to the possibility of such leaks, lifelong monitoring, typically using CT scans with

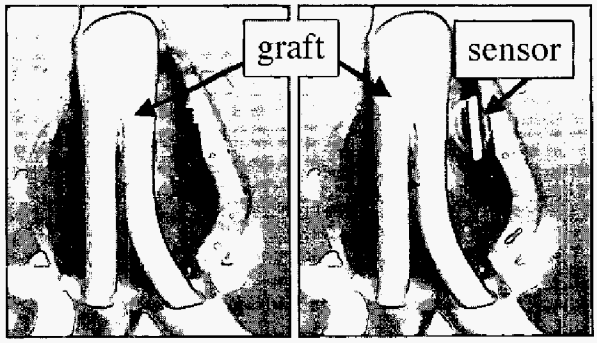


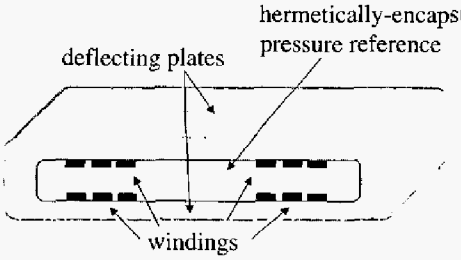
Figure 2. *Left: Endovascular stent-graft repair of AAA. Right: Sensor incorporated into AAA sac to monitor sac pressure.*

potentially nephrotoxic contrast dye, is indicated for the rest of the patient's life. The repeated use of this dye has been associated with reduced kidney function in persons undergoing endovascular repair when compared with surgical repair [16]. A better monitoring concept would be to place a permanently-implantable wireless pressure sensor in the aneurysm sac during the endovascular implantation of the stent graft. This sensor monitors the intrasac pressure and is interrogated using external RF electronics in a noninvasive fashion, without the need for dye or radiation (Figure 2, right).

APPROACH

The general sensor concept is shown in Figure 3. Flexible plates bearing inductor windings (along with associated distributed capacitances) bound a hermetically-sealed reference cavity. The inductor windings serve two purposes: to form a resonant electrical circuit with the capacitor; and to magnetically couple with an external loop. A change in the pressure surrounding the sensor will change the position of the plates, thereby changing the capacitance and resonant frequency of the sensor. The change in resonant frequency will cause a change in the response of the external loop, which can be monitored by external electronics and through appropriate calibration be translated back into the pressure sensed by the sensor.

In order to ensure stability of the sensor, it is necessary to utilize a hermetically-encapsulated pressure reference; to ensure that the majority of the energy stored in the sensor does not penetrate into the lossy and potentially variable dielectric medium in which the sensor is embedded; and to ensure that the compliance of the plates is small when compared to the compliance of any material (e.g., clot) that might form on the sensor surface. The use of MEMS fabrication techniques enables very stiff plates while simultaneously maintaining relatively large pressure sensitivity (approximately −10 kHz/mm Hg) by the realization of small, stable gaps.



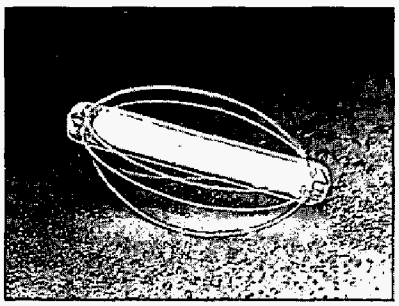


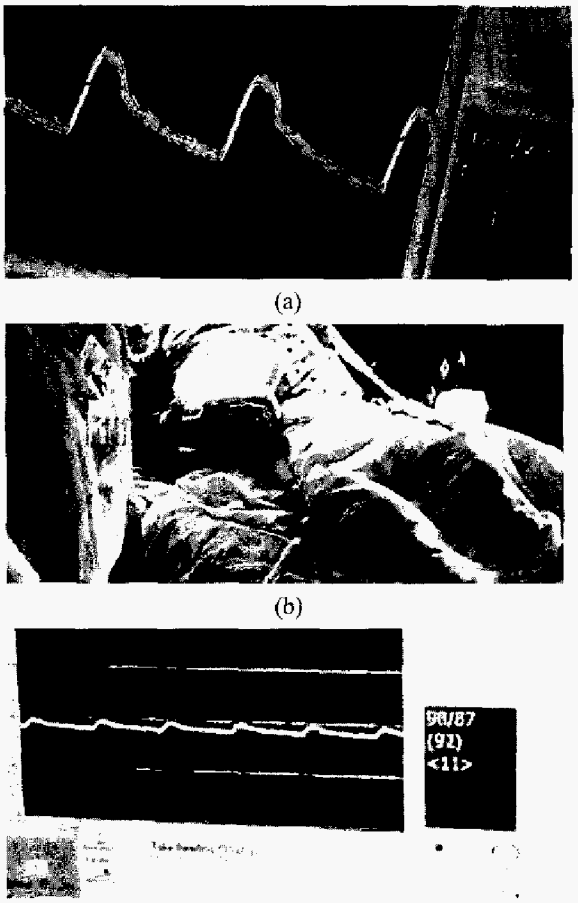
Figure 3. Top: Schematic cross-section of sensor; bottom: fabricated sensor. The 'basket' surrounding the sensor has no electrical functionality but acts to keep the sensor anchored within the sac. The sensor is approximately 5mm in width and 30mm in length so as to sample a reasonable fraction of the large volume of the aneurysm sac.

Sensor fabrication proceeds using standard MEMS approaches. A lower substrate of fused silica less than 1mm in thickness has the lower inductor electrodeposited on it using a plate-through-photoresist-mold approach. An upper substrate of fused silica has a recess etched into it using isotropic wet etching and the upper inductor is electrodeposited into the recess using the same plate-through-mold approach. The two substrates are then fusion-bonded. Careful control of the electrodeposition and recess etching processes allow the gap between upper and lower inductor windings to be controlled with micron-order precision. Note that to enhance reliability there are no electrical interconnections in this design. Typical resonant frequencies of sensors are in the 30-40 MHz range with Q-factors of approximately 50 and readout distances of approximately 8 inches. Similar readout distances have been achieved with circular sensors as small as 3.8mm in diameter.

The sensor is introduced through the femoral artery contralateral to the side in which the main endograft will be inserted, using the same sized sheath (14 Fr) that is used for introduction of the contralateral limb of the graft itself. During subsequent introduction and deployment of the stent-graft system, the sensor is maintained in position using a tethering system. Once the stent-graft and the contralateral limb are deployed, the sensor is released from its tether and the tether system is removed, leaving the sensor permanently inside the sac. Sensor interrogation results in a real-time pressure trace of the sac pressure, which is displayed on an external monitor.

RESULTS AND DISCUSSION

Sensors have shown excellent correlation with both wired sensors in an animal model [17] as well as with intraoperative pressure catheters placed in the aneurysm sac. Typical drops in pulsatile pressure exceeding 30% are observed upon exclusion of the aneurysm sac, with concomitant long-term drops in mean pressure when endografts are successfully sealed. Figure 4 shows typical real-time pressure traces obtained by the sensor at various stages during the implant procedure and follow-up. Note that it is possible for the patient to read his own sac pressure. To date, approximately 100 patients have received sensor implants at the time of endovascular repair of their AAA, with the longest implant times exceeding 1 year (all sensors functional).



(a)

(b)

(c)

Figure 4. *Sensor operation. (a) Pressure trace of systemic pressure prior to sac exclusion; (b) patient taking own measurement at 24-hour followup; (c) sac pressure at 24-hour followup (note reduced pulsatility indicating sac exclusion).*

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