The Role of MEMS in Endovascular Pressure Sensors


Niels Black
Jonathan Brickey
Charles Wang
Northwestern University
Prof. Horatio Espinosa
December 9, 2007
Table of Contents

I) Introduction ................................................................................................................................. 3

II) Project Description ..................................................................................................................... 5
   i) Literature Review ....................................................................................................................... 5
   ii) Why MEMS are an important part of this device ................................................................. 12
   iii) The Device Itself .................................................................................................................... 13
       Operation .................................................................................................................................. 13
       Capacitance-Pressure Relationship ....................................................................................... 13
       Inductance ............................................................................................................................... 15
       Circuit Resonance .................................................................................................................... 15
       RF Transmission ...................................................................................................................... 16
       A Note on Simplification: ........................................................................................................ 18
   iv) Future Improvements ............................................................................................................. 19

III) References ................................................................................................................................ 22

IV) Biographical Information ......................................................................................................... 24
I) Introduction

The heart pumps oxygenated blood out of its left ventricle to the rest of the body via the aorta. The abdominal aorta is the largest artery in the abdominal cavity, and supplies blood to most of the lower body. In order to deliver enough blood in a timely fashion, blood pressures in the abdominal aorta typically range from under 80 mmHg (~11 kPa) during diastole to under 120 mmHg (~16 kPa) during systole for healthy adults. To accommodate this repeated change in pressure, the aorta walls are highly elastic. However, as a person’s aorta ages, it loses elasticity and distensibility, and begins to bulge (aneurysm) at the weakest region due to the high blood pressure (systolic pressure greater than 140 mmHg (~19 kPa) for people with hypertension) in the aorta (www.medicinenet.com). The vessel wall thins and eventually ruptures, causing internal hemorrhaging and death if not promptly treated.

Abdominal aortic aneurysm (AAA) is the tenth leading cause of death in the U.S. for people aged 65-74, as 1-3% of men over 65 experience an aortic rupture, the mortality rate of which is 90% (www.familypracticenotebook.com and Lindholt). Therefore, early detection and treatment are crucial, but this is difficult because 75% of AAA are asymptomatic. The outer diameter of a healthy aorta is about 2 cm, with diameters over 3 cm as indicators for aneurysms (family practice notebook). Two types of treatment are generally available: open and endovascular repair. The former is usually more suitable for younger patients or those with rupture aneurysms. Endovascular aortic repair is usually performed (when risk of rupture becomes greater than surgical risks, usually when the diameter of the aorta reaches 5.5 cm), by inserting a stent graft from the femoral arteries and guiding it to a position that reinforces the weakened aortic sac (Rutherford). An internal blood pressure sensor can be attached to the graft wall and
left inside the body to take real-time measurements during post-surgical recovery and subsequent checkups.

The EndoSure Wireless AAA Pressure Measurement System (by CardioMEMS) claims to be the first commercial, un-powered, permanently implanted blood pressure sensor that combines MEMS technology and wireless communication (Becker). Only the size of a paper clip, this device’s biocompatible, elastic enclosure deforms when blood flows through the vessel (stent graft) to which it is attached. It is thus able to “measure” blood pressure constantly. However, the measurements are only transmitted when the external controller is brought within range and is able to power the implanted device. Current alternative methods include imaging techniques, such as the CT scan, that are much more expensive, only shows the relative sizes of the bulge (not actual pressures), and cannot provide real-time data in emergencies.

Microfabrication techniques are needed to achieve the size of this device, including its two major components: the pressure sensor and radio frequency transceiver. Because this product is currently on the market, details of the fabrication procedure and exact mechanisms of the different parts are not known. Therefore we plan to examine two similar pressure sensors developed in the same lab in the previous years: a ceramic high-temperature sensor and a flexible biocompatible sensor and from there speculate how the EndoSure sensor likely function and is fabricated.
II) Project Description

i) Literature Review


In 1967, C. C. Collins created the first miniature passive pressure transducers. They were manufactured with macro machining operations in diameters of 6, 4, and 2 mm. Two connected hand-wound coils were attached to Kevlar membranes stretched over a glass ring and coated in polyethylene. They were produced for implanting in the eye and required an external powering and detecting device at most 3cm away. The stiffness of the membrane had a natural frequency several orders of magnitude larger than that of the LC coils so that it did not affect the pressure sensing or show any acceleration effects. The thickness of the diaphragm and the separation distance of the coils could give a range of different effects. A thin diaphragm and closely spaced coils were very accurate (100 MHz/mmHg), but low pressure range due to the small gap and a shorter life due to the permeability of the membrane. The devices also experiences plastic deformation for 3 weeks before the device is precise.
It wasn't until 1992 when a Swedish team created an LC pressure sensor for intra-ocular pressure detection using MEMS fabrication techniques when make the variable capacitor. Silicon was used for the substrate and diaphragm and small external inductor was attached to complete the LC circuit. It also only operated in very short ranges, up to 22mm. It produced a very good relation between pressure and resonant frequency, but had a very low sensitivity (2 kHz/mmHg). It also had a low Q-factor due to the large external inductor's resistance and stray capacitance due to SiO2 layers that formed on the silicon surfaces.

Then in 1995 a New Jersey Institute of Technology team used MEMS techniques to create a pressure sensor with integrated hi-precision inductor coils. It also used a Hartley oscillator setup to separate the inductor for the resonating LC circuit and the transmitting/receiving inductor. It provided a much higher sensitivity of 2.5 MHz/mmHg at resonate frequencies of 50-90 MHz.

Jay S. Yadav, M.D and Mark G. Allen, Ph.D founded CardioMEMS in late 1999. The company was created for the purpose of creating a biocompatible pressure sensor.
Earlier that year, Mark Allen had created a ceramic wireless pressure sensor for use in turbines. It used layers of ceramic tape to form a capacitive chamber, photo masking and copper electroplating on the bottom and DC sputtering deposits titanium/copper layers on the top to create the spiral capacitive/inductive plates of a passive LC resonator. It showed little temperature dependence. Its sensitivity was rather good at 2 GHz/mmHg. The two coils touch at high pressure, reducing sensitivity, and the device is not self-packaged.

In 2001 Mark Allen made a second ceramic pressure sensor. The previous sensor used a glass ceramic that became conductive at high temperatures. This one used alumina ceramic and fully encapsulated the internal silver LC circuit. It could withstand much higher temperatures and
pressures, though it was less sensitive, showing an accuracy of around 52-180 MHz/mmHg at 43 MHz.


A 2005 report by mark Allen gave the world a first look at the Endosure device CardioMEMS was working on. By using precision MEMS fabrication and stiff diaphragms, the windings can be placed very close to each other, resulting in sensitivities of 10 kHz/mmHg and Q-factor of 50. The simplicity of the device leads to increased stability and reliability and low manufacturing cost. With this setup and coils as small as 3.8 mm in diameter, detection distances work as far away as 8 in.
In parallel with the Endosure, Mark Allen and others at Georgia Tech have been working on similar implantable wireless pressure transducers. In 2006 they reported on 2 different flexible designs. One printed on flexible liquid crystal polymer and one with a ceramic reference pressure chamber. The devices had diameters of 11mm and sensitivities characterized between -1 to -20 kHz/mmHg for the LCP and 0.5 to -1.5 kHz/mmHg for the Ceramic. Although easy to manufacture, they suffered from short life spans due to the permeability of their membranes and inconsistent results due to the change in the capacitance resulting from changes in the dielectric constant of the surrounding environment.
On April 9, 2007 the FDA approved the use of the Endosure device. The device comes in 3 sizes: 3.2cm, 4.1cm and 5cm. The device shows promising patient results and should be expected to enter the market soon.

**Manufacturing Processes**

Much attention is being paid to the field of flexible electronics. Obviously, a flexible substrate is very preferable for tight, sensitive conditions, such as the interior of the body, but it also provides excellent shock protection. Several common substrates used for modern applications include polyamide films (notably Kapton-E from Dupont) and various types of liquid crystal polymers. The most important factor for these substrates is dimensional stability, something polymers are not typically known for. With the tiny distances measured by MEMS in general, even a dimension change as small at a few tenths of a micron can have a huge impact on the functionality of the device. While these substrates are flexible, they deform very uniformly in each direction to keep distances equivalent. One possible disadvantage is the strong isotropic behavior of polymers. In some cases, this means etching is not the best practice, since while chemical etching works very well, it can be difficult to obtain the desired structure.

MEMS screen-printing operates on principles very similar to cloth screen-printing. The first step is to paint a negative image of the desired structure onto a mesh. This mesh (typically made of stainless steel or polyester) is placed over the substrate, and the material to form the
structure is passed over it in liquid form. The mesh spacing is typically on the order of 50 µm. Additional structures can be placed on the substrate by different mesh masks. Micro-screen-printing, like its macro relative, is relatively cheap, and is excellent for mass production. The mesh masks are more durable than traditional photolithography masks, although they are generally subjected to higher stresses. Screen-printing has the advantage of not requiring pressurization like lithography, and is excellent for printing on flexible substrates, since etching is not required. However, it cannot achieve the high resolution of lithographic methods.

Optical lithography is the standard for MEMS fabrication. On a basic level, it involves the use of a photoresistive film, combined with a mask in the desired shape of the pattern. UV light is shined through the mask onto the film, which will either become harder or softer, depending on the type. The soft photoresist is then washed away through chemical processes. The exposed substrate can then be etched using whichever process produced the desired results. Using this method, smaller feature resolution than screen-printing is achieved, but etching on flexible substrates can have unpredictable results. One method which meets with success is etching of channels on the substrate which are then filled with conductor. Copper is an excellent choice for this application, since Kapton-E polyamide film has an identical thermal expansion coefficient.

While laser fabrication methods are not MEMS-specific processes, they are important for post-processing. The actual MEMS component of the sensor has features on the order of tens of microns, but the outer casing is significantly larger. While it is certainly possible to manually manipulate the sensors, it is far more economical for mass production to use lasers. Several different methods are used, depending on the application:
CO₂ laser – 10 µm wavelength – The jack-of-all-trades CO₂ laser is standard equipment in most labs due to its excellent efficiency. It is generally used for relatively coarse work, such as separating each individual sensor from a sheet of several.

Excimer laser – 248 nm wavelength – This laser is excellent for fine polymer work, both due to its small wavelength and because it pulses rapidly enough to remove material without significant heating or melting.

ii) Why MEMS are an important part of this device

The main problems associated with body-implantable sensors are size and accessibility. Obviously, it’s far better to have a device the size of a paperclip next to your heart than one the size of a stapler. The initial implantation is far less stressful on the body, particularly if the device can be implanted via catheter, rather than open-chest or endoscopic surgery. Accessibility is important for both getting the data from the device and providing power for operation. MEMS devices are beneficial for both of these: by definition they are far smaller than components obtained using more traditional methods (machining, etc.) and this small size means a very small amount of power is needed. This power is small enough that it can be easily provided by an external means, and with careful and ingenious design, the data can be gathered through similar processes. This eliminates the need for wires leading into or out of the body, wires which are both inconvenient for the patient and an avenue for potentially lethal infection.
iii) The Device Itself

Operation

At the heart of the EndoSure (and most capacitive pressure sensors) is a cavity bounded on two sides by a thin, flexible membrane. Embedded in the membrane are micro-scale conducting components, forming a capacitor. As the pressure around the sensor changes, the membrane deflects in a predictable relationship. This deflection changes the capacitance, and when combined with an inductor, this is reflected by a change in the resonant frequency of the circuit.

Capacitance-Pressure Relationship

Determine precise capacitance can be somewhat challenging, but fortunately it can be simplified when \( d^2 \) is much less than \( A \) to \( C = \frac{d^2}{d} \).

This is a simple relationship between the space between the capacitor plates and the capacitance. Unfortunately, this is assuming the capacitor plates remain parallel to each other, whereas flexible membranes will bulge rather than move as a plane (Figure 1).

![Figure 1: Cross section of a circular capacitor plate under constant pressure (Espinosa)](image)

To find the change in capacitance then, it is necessary to perform an area integral over the plates:
Assuming a circular capacitor, the bulge will be uniform in polar coordinates, and can be greatly simplified as such:

$$C = \int_{r} \left[ \frac{\varepsilon_0}{d_0 - w(x,y)} \right] dA$$

It is important to note that since the capacitor in question is comprised of two flexible membranes, the value of $w(r)$ is doubled to account for this decreased plate separation. According to engineering formulas for edge-constrained beams subjected to a uniform force, the deflection of the beam will be equal to:

$$w(r) = \frac{3(1-\nu^2)}{16Eh^2} \frac{p_o}{r_0^2 - r^2}$$

Where $E$ equals the elastic modulus of the material, $\nu$ is poisson’s ratio, $p$ is the pressure, $h$ is the initial separation of the plates, and $r_0$ and $r$ represent the total radius and the radius in question. Integrating over the area of the capacitor gives

$$C(\Delta p) = \pi \varepsilon_0 / d_0 [r_0^2 - \Delta p(1-\nu^2) r_0^4 / (8Eh_0^2)]$$

This is the relationship between pressure and change in capacitance. While integrating using these approximations, we find that the capacitor shorts out at pressures just under 16 kPa, which is around the blood pressure in a hypertensive aorta. The literature mentions that the actual capacitor shape is tapered in the center to reduce deflection and prevent “shorting out the
capacitor over pressure excursions of interest.” (Fonseca 2006) Clearly, improvements need to be made to the model in order for these approximations to be used.

**Inductance**

The inductance of a planar spiral inductor can be roughly determined by

\[ L = \mu_0 n^2 r = 1.2 \times 10^{-6} n^2 r \]

Where \( n \) is the number of turns in the capacitor and \( r \) in the radius. Plugging in the dimensions of the Endosure predecessor (12 turns, 3.8 \times 10^{-3} \text{ m}) gives approximate inductance of 0.656 \( \mu \text{H} \).

**Circuit Resonance**

The resonant frequency of an LC circuit is calculated as

\[ f_0 = \frac{1}{2\pi \sqrt{LC}} \]

In this circuit, the impedance remains constant while the capacitance changes. Having found the relationship between pressure and capacitance in the previous section, it can easily be shown that:

\[ \Delta f_0 = \frac{1}{(2\pi \sqrt{L})} \times (1/\sqrt{C(p_1)} - 1/\sqrt{C(p_2)}) \]

Where \( C(p) = \pi \varepsilon_0 / d_0 [r_0^2 - p(1-v^2)r_0^4/(8Ed_0^2)] \)

**Resistance:**

\[ R = \frac{l \cdot \rho}{A} \]

Resistance is estimated at \( 1.7 \times 10^{-8} \Omega \text{m} \times 0.286 \text{ m} / 1.08 \times 10^{-3} \text{ m}^2 = 5 \text{ micro-} \Omega \)
**Q-Factor:**

\[
Q = \frac{1}{R} \sqrt{\frac{L}{C}}
\]

At \( p = 11 \text{ kPa} \), the capacitance is calculated as 1.45 pF. Plugging the aforementioned values into this equation yields a Q factor of \( 1.64 \times 10^6 \).

**RF Transmission**

![Equivalent circuit of electrical model for a sensor coupled with an antenna (Fonseca 2002)](image)

When the external controller is powered on, a voltage \( V_1 \) is generated and the current \( I_1 \) that flows in the antenna loop produces a magnetic field around inductor \( L_a \). The current is AC with a designated frequency content. When the controller is brought within range of the implanted sensor, this magnetic field then induces a current in \( L_S \) and powers the device. And the voltage across \( L_S \) is:

\[
V_2 = L_1 \frac{dI_1}{dt} + L_M \frac{dI_2}{dt}
\]
The inductor begins to store energy. When it is fully “charged” after some time, the controller switches mode from active to passive “listening.” In the sensor circuit, the induced frequency will eventually decay to the resonant frequency $f_o$, as defined by the inductance $L_s$ and capacitance $C_s$, which is a function of the changing pressure of the abdominal aorta. This frequency can be detected as an impedance in the antenna:

$$Z_i = \frac{V_i}{I_i} = j2\pi f L_s \left[ \frac{\left( \frac{f}{f_o} \right)^2}{1 + k^2 \left( \frac{f}{f_o} \right)^2} \right]$$

\[\frac{1}{1 - \left( \frac{f}{f_o} \right)^2 + \frac{1}{Q f f_o}}\]

(Fonseca 2002)

By comparing the measured impedance to the original calibration (representing healthy blood pressure), it is possible to detect abnormalities. (English) After some signal processing to interpret the changes in magnitude and phase of the impedance, a real time pressure readout can be generated.
This figure compares the reference sensor, which is a wired device, to the wireless device’s measurements, both of which are implanted into a dog.

*A Note on Simplification:*

An important consideration in this analysis is the simplification of the system. Calculating the actual inductance, resistivity, and capacitance would take an inordinate amount of time and resources, so approximations are used. These drastically affect the outcome of the modeling, making some aspects of the Endosure impossible to calculate.
iv) Future Improvements

When implanting a device in the body, size is a big concern. The Endosure device is relatively large compared to its predecessors, as it is in the range of several cm rather than mm. This is due partially to over design to assure reliability, but mostly in order to increase the distance the sensors can be read by the external monitor.

To understand the limiting factor in size of the sensors, we need to start with an approximation of the inductance of the resonating coil. For a flat, round, spiral coil printed on a substrate, this can be estimated as

\[ L = N r \left( \ln \frac{8r}{a} - 2 + Y \right) \]

where \( N \) is the number of turns, \( r \) is the average radius of the coils, \( a \) is the wire radius, and \( Y \) is estimated to be between 0 and \( \frac{1}{4} \) depending on the electron density. For a flat, rectangular, spiral coil, the inductance is approximated to be

\[ L = N \frac{1}{\pi} \left( b \cdot \ln \frac{2b}{a} + d \cdot \ln \frac{2d}{a} - (b + d)(2 - Y) + 2 \sqrt{b^2 + d^2} - b \sinh^{-1} \left( \frac{b}{d} \right) - d \sinh^{-1} \left( \frac{d}{b} \right) \right) \]

where \( b \) and \( d \) are the average lengths of the two sides of the rectangle.

Using the equations for magnetic flux we are able to relate the inductance and the magnetic field inside the coils of an inductor. \( i \) is the current, \( L \) the inductance, \( B \) the magnetic field, and \( A \) cross sectional area of the coils.

\[ \phi = Li \quad \phi = BA \]
From the Boit-Savart law we can calculate the magnetic field a distance $z$ perpendicular from the plane of a circular inductor.

$$B(z) = \frac{\mu_0 i r^2}{2 \left(r^2 + z^2\right)^{3/2}}$$

Knowing the magnetic field at $z=0$, we can substitute the inductance of the coil to get the relation:

$$B(z) = \frac{Li r}{\pi \left(r^2 + z^2\right)^{3/2}}$$

From this equation we can see that the magnetic field is directly related to the inductance of the resonating coil, and if $z \gg r$, then the magnetic field is inversely related to the distance cubed. Because we are working at very low currents and sizes, the Magnetic field is not very strong to begin with, and drops off quickly. Most magnetic field detectors can only register several gauss.

There are other ways of increasing the inductance of a coil other than just increasing the side. Inductor design can be very complicated, and there are several computer programs for calculating the inductance of a coil. As reported by Collins, connecting the two capacitive coils can increase their capacitance due to their mutual inductance. Also, placing several coils intertwined or on top of or next to each other can give various improvements.

The recent development of low power nanopower low output impedance op-amps by R. Drescher and P. Irazoqui make the use of other more complicated oscillating circuits possible. William N. Carr demonstrated the improvements of using a Hartley oscillator in a similar pressure sensor. Other improvements can be made with RF to DC converters. At the sacrifice of
fewer pressure readings per second the sensor could store energy for amplifying the current for signaling the pressure, greatly increasing the range of the device. It may even store enough energy to send out a warning signal to a cell phone or similar device carried with the patient to let them know if a critical pressure was observed.

Another problem is the dropping of the resonant frequency and quality factor when in a saline solution as opposed to air. This is due to the saline solution having higher electromagnetic losses as compared to air. The field extending from the inductor coil loses energy much more readily in the saline solution. It also changes the capacitive value at a given pressure if the sensor is against the blood vessel, the stent graft, or just in the blood flow.

The Endosure uses several wire loops to keep the device from falling against any surfaces, but increases the overall size of the sensor and adds the danger of the wires dislodging. Some suggested solutions for this problem have been to coat the device with silicone or other material with a low dielectric constant to shield it from the saline.
III) References

“Abdominal Aortic Aneurysm.” FamilyPracticeNotebook.com
<http://www.fpnotebook.com/SUR2.htm>


Espinosa, H. D., “Lecture on MEMS pressure sensors” <https://courses.northwestern.edu/@@DB34E5673F5154CDB77244DEC396BC70/courses/1/2007FA_MECH_ENG_381-0_SEC20/content/_1204461_1/lecture5-pressure-sensor.ppt>


“High Blood Pressure (Hypertension).” MedicineNet.com <http://www.medicinenet.com/high_blood_pressure/article.htm>


IV) Biographical Information

Niels Black is a senior at Northwestern University’s McCormick School of Engineering and Applied Science, where he is majoring in Mechanical Engineering. He is a half Dane from South Dakota with an interest in design and robotics. He enjoys the challenge of new projects and dreams of the future. He also enjoys singing, gaming, and computers.

Jonathan “Bob” Brickey is a junior at Northwestern University’s McCormick School of Engineering and Applied Science, where he is majoring in Material Science. A native of Colorado, he is the youngest of three. He developed his interest in material science from his brother, who attended Rice University and is now an engineer with the Colorado Department of Transportation.

Charles Wang is a first year graduate student at Northwestern University’s McCormick School of Engineering and Applied Science, pursuing a M.S. in Biomedical Engineering. He received his B.S. in Biomedical and Electrical Engineering from Duke University in 2007. Charles’s research interests include neural prosthetic devices and human lower limb gaits. He hopes to begin work on his thesis project at the Rehabilitation Institute of Chicago in the spring of 2008. Charles plays and watches basketball in his free time, enjoys volunteering as a tutor, and is involved with Engineering World Health projects. He hopes to start up and run his own company in the future.