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1 Overview

The Micro-Electro-Mechanical Systems (MEMS) have found wide application in medicine these days. Lots of them are used in form of implanted devices which usually demand complicated design and fabrication requirements. One of the most important issues for implantable MEMS is to be biocompatible. A conformal coating can satisfy the requirements for medical surface modification. Such coating can isolate the surface from body fluids and tissue, insulate it electrically, provide enough lubrication for the system, stabilize the surface against immobilization of microparticles, limit the contaminants entrapment by its nonporous surface, improve mechanical properties of the system and promote the growth of tissue [68, 81].

This work aims to evaluate different biocompatible coatings on an implantable MEMS pressure sensor.

An implant usually needs a coating which can withstand the aggressive body fluid as well as being biocompatible. The coating must protect the sensor in the environment of the body. Meanwhile, it should not disturb the implant performance or damage the host tissue.

Coating might change the sensitivity and offset in an implantable MEMS sensor. In this project variation of mechanical and thermal characteristics of pressure sensors before and after coating are studied. These characteristics mainly include the sensitivity and the temperature coefficient of the offset.
2 Why MEMS

Today, integrated circuit fabrication technologies are used to produce Microelectromechanical systems (MEMS) in miniature scale, and in great quantities. In addition to small size, MEMS-based sensors have lower weight, higher reliability, lower power consumption, higher accuracy, better performance, and cheaper mass production [1, 2, 6, 11, 14, 22].

Silicon MEMS sensors have higher sensitivity, uniform performance, and lower mass production cost compared to other technologies such as piezoelectric materials [1].

2.1 Why MEMS in Medicine?

The aforementioned advantages, combined with the lower risk of infection leading to a low risk of mortality in intensive care [22], are good enough motivations for extensive use of MEMS in medicine. MEMS system technologies (or microstructure technology), MST, provide outstanding opportunities and are expected to have a significant influence on many areas including medicine [1, 11].

2.2 Why implanted Pressure sensor

Any device working inside the body either for short or long term is considered an implant. Implantable MEMS sensors are, for instance, utilized to aid diagnosis or to provide a means of feedback control for therapy [11].

Producing electrical and mechanical systems were the first applications of semiconductors. These applications are now enhanced to other systems such as biological, optical, magnetic and fluidic systems [1].

A microelectromechanical sensor mainly consists of a control unit, a sensing and actuating unit and a power supply. These parts are shown in Figure 2-1 schematically. MEMS can integrate four different essential functions in one miniature unit. These systems perform both sensing and actuating as well as computation. They are also able to communicate with locally controlled physical parameters [1].
Implanted MEMS sensors have had strong impact in the medical world and paved new ways for monitoring and therapy and improved health care services [2]. There are lots of different kinds of implantable Microelectronic systems on the market. The most commonly used device is the cardiac pacemaker which has been implanted around 600000 times around the world. The next most frequently used is the defibrillator [3]. Among sensor devices piezoresistive sensors are the most commercially available implantable sensors [14], [28] (p.469).

Measuring temperature and pressure are among the most frequent measurements in clinical diagnostics and therapies. [3, 4]

Some of common examples of medical pressure measurements can be listed as

- blood pressure,
- intraocular pressure,
- bladder pressure,
- pressure within big joints
- abdominal aortic aneurysms (AAA)
- Intracranial pressure

Work on fully implantable blood pressure sensors [4, 5] and intraocular pressure sensors [6] are in progress [2].
Thanks to fully implanted Microsystems it is now possible to have continuous monitoring of pulse rate, blood pressure or intracranial pressure [5].

In the rest of this chapter several pressure sensor solutions will be described.

### 2.2.1 Blood pressure sensor

Some examples of intravascular pressure sensors are highlighted below

a) *Fiber optic:* Fiber optic is a promising solution for the invasive measurement mainly because of the biocompatibility of $\text{SiO}_2$, electrical passivity and electromagnetic interference (EMI) immunity. Usually a reflective diaphragm is used at the end of optical fibers in fiber optic pressure sensors. Two detection principles can be applied in these reflective diaphragms. The first one is an intensity-based technique [38], in which the reflected intensity is directly proportional to diaphragm deflection. This method has the advantage of simple processing scheme; however, it has fluctuations in its optical fiber and does not provide high sensitivity. The second principle is an interferometric technique that in general provides the highest resolution. This principle is based on detecting a phase change of the light reflected from the diaphragm [4].

Edvard Cibula *et al* have worked on a fiber optic pressure sensor for invasive blood pressure measurements. Its diameter is only 125µm. Being cheap, biocompatible, ecologically acceptable, applicable for wide range of pressures and simplicity of fabrication process are some advantages of this sensor [4]. Sensing element in this pressure sensor is a Fabry-Pérot interferometer[^1] (see Figure 2-2), FPI, which is placed at the tip of an optic fiber (Figure 2-3).

[^1]: “In optics, a Fabry-Pérot interferometer or etalon is typically made of a transparent plate with two reflecting surfaces, or two parallel highly reflecting mirrors” [13]
**Figure 2-2 [13]:** When the light enters an FPI it will be reflected several times. Here $T_i$ is the first transmitted beam, $\theta_i$ is the emission angle, $n$ is the reflective index of the material between the reflecting surfaces, $R_1$ is the first reflected beam and $l$ is the distance of two reflecting mirrors. The phase difference between each transmitting and reflecting beam can be calculated for known wavelength.

**Figure 2-3 [4]:** Sensing element of fiber optic pressure sensor

Reflecting polymer diaphragm and inner fiber cavity interface are the main parts of the interferometer. Light emitted by a broadband source (LED) which comes through a 2×2 coupler, interferes inside the FPI. The 2×2 coupler or four-port directional coupler is the simplest optical coupler (see Figure 2-4 (a)). When the light arrives for example at A, it splits between port C and D. The light has eight possible ways to travel in such coupler (Figure 2-4 (b)) [39]. The wavelengths that satisfy the condition of constructive interference are reflected back from FPI through the 2×2 coupler to the spectrometer. The FPI behaves like an optical filter whose spectral properties depend on cavity length. When an applied pressure deflects the elastic polymer diaphragm, the length of the cavity is modified, and detected in the spectrometer by a shift in the reflectance spectrum.

**Figure 2-4:** (a) A schematic of a 2×2 coupler. (b) Overall, there are eight possible ways for the light to travel in a 2×2 coupler
b) *Cuff wound around the blood vessel:* In this type of pressure sensors the blood vessel is surrounded by a biocompatible cuff [3, 12]. Peng Cong *et al.* used soft silicon to make this cuff because of its biocompatibility as well as having enough elasticity. This cuff is filled with silicon oil and the pressure sensor is submerged into it (Figure 2-5). As the cuff is soft and elastic, its stiffness can be smaller than that of a blood vessel. So the restrictive effect on the vessel is generally minimized.

![Figure 2-5](image)

**Figure 2-5 [40]:** Cuff around a blood vessel with a pressure sensor integrated in it

The cuff has a close contact with the vessel for more accurate pressure sensing. It is better to have a pre-curved cuff structure so that is possible to wrap it around the vessel during implantation (Figure 2-6). This can also have a uniform contact to the vessel wall. The cuff structure is a curved silicon membrane attached to a silicon base. The base's dimension is 5mm×2mm×100µm, suitable for wrapping around an animal vessel with approximately 1mm diameter. The prototype base has 7 sets of bump supports with 100µm height. These bumps support the membrane to avoid touching the bottom of the silicon base.

![Figure 2-6](image)

**Figure 2-6 [12]:** The cuff's curved structure

Then the cuff is filled with medical-grade silicone fluid through the channel opening shown in Figure 2-6. The MEMS pressure sensor will be immersed into the fluid and
the cuff will be sealed. The blood pressure, $P_b$, can be expressed by equation 2-1 where $P_m$ is monitoring cuff pressure, $P_{bias}$ is the cuff bias pressure and $\mu$ is a scaling factor.

**Equation 2-1**  
$$P_b(t) = \frac{1}{\mu} (P_m(t) - P_{bias})$$

$P_b$ and $P_m$ are functions of time. $P_{bias}$ and $\mu$ can be obtained by using a catheter tip or tail-cuff device [40]. They may change slowly over time and can be calibrated by means of systolic and diastolic blood pressure levels. The pressure sensor explained in this section is tested *in vivo* in the right carotid artery of a laboratory rat. *In vivo* results are in accordance with the results from catheter-tip method with a constant scaling factor [12, 40]. This method bypasses vessel penetration and minimizes vessel constriction because of the soft cuff elasticity. Therefore it is interesting to avoid long-term adverse biological effect compared to the conventional catheter-tip implant technique, the most common technique in monitoring blood pressure in small animals.

Babak Ziaie *et al* worked on another configuration. It consists of a sensor array with a titanium support cuff for tonometric blood pressure measurement in small animals [14]. Tonometry is a non-invasive technique for continuous pressure measurement in closed vessels such as blood vessels, bladder, intraocular and brain pressure. Using this technique for intraocular pressure measurement has been successful [41, 42] while blood pressure measurement had a limited success [43]. To measure the blood pressure, the vessel is pressed flush against the sensor’s membrane. According to the Laplace’s law, it can be proved if the vessel is flattened enough; the pressure inside the vessel is equal to the measured pressure (See equation 2-2, where $P_{in}$ is the inside pressure, $P_{out}$ is the outside pressure, $T$ is the tension of the vessel wall and $r$ is the radius of the vessel.)

**Equation 2-2:**  
$$P_{in} - P_{out} = \frac{T}{r}$$
In this equation it is assumed that the vessel radius is much larger than the wall thickness (one order of magnitude). When \( r \rightarrow \infty \), the vessel is completely flattened, \( P_{in} - P_{out} \rightarrow 0 \), the measured pressure is equal to blood pressure. There are three important requirements in this measurement technique. Firstly flattening the vessel should not cause occlusion or significant hemodynamic disturbance. Secondly the stiffness of the pressure sensor diaphragm should be greater than that of the vessel wall to ensure appropriate flattening and avoid excessive bending of the vessel wall during measurement. Thirdly the sensor's membrane should be smaller than the artery. External arterial tonometers which are placed on wrist have not been commercially successful because of their inaccuracies. Imprecise positioning of the sensor and degree of arterial flattening (excessive wrist movements) results in inaccurate measurements. Limited movement of implantable tonometers makes them more immune to these sources of error. More precise sensor placement is possible by means of micromachining technologies [44]. In the system explained by Ziaie and coworkers an array of capacitive MEMS pressure sensors with a titanium cuff have been used (Figure 2-7). The system is designed to measure arterial blood pressure in small mammals (e.g. rat). An array of three pressure sensors is placed underneath the vessel to increase stability. A rectangular diaphragm has greater central deflection compared to circular or square diaphragms with similar size. So the sensor with rectangular diaphragm would be more sensitive. The sensor explained is designed to have a long rectangular shape diaphragm.

![Figure 2-7](image)

**Figure 2-7 [15]:** Implantable tonometric pressure blood pressure sensor
The titanium cuff has two parts: a base and a cap as shown in Figure 2-7. The titanium base supports the sensor array and interface chip. The cap is screwed to the base by means of miniature screws. The pressure sensor array is attached to a rectangular recess by means of silicon rubber to be fixed during the operation. In this implantable pressure sensor the package can easily induce stress which will be transferred to the transducer. This is the main source of base-line drift in such sensors. Consequently, the assembly and packaging of the system is vital to decrease the drift. The blood vessel is positioned on top of the sensor array. Then the cap is screwed to the base to clamp the vessel. The vessel is pressed to flatten against the sensor diaphragm. The system has been tested in vitro. Results show sensitivity of $2\text{mV/mmHg}$ and a resolution of 0.5mmHg at 100mmHg.

c) **Telemetric low power:** Implantable microsystems can be categorized into two main groups: battery driven systems and systems without internal power source. Limited battery lifetime is the main disadvantage of the first group. The second group should have big enough antenna in order to provide sufficient power. Therefore size reduction is the limitation for these systems. By separating the sensor and telemetry it is possible to design much smaller sensor elements with larger telemetric part. In this case, the sensor can be implanted in the measurement spot whereas the telemetry part is placed directly under the skin. This optimizes the energy transfer [45]. In addition to the sensor’s size reduction, the other advantage of separating sensor and telemetry part by having a modular system concept is the possibility of having different sensors for diverse applications while having only one telemetric unit [5].

Schlierf *et al* presented a sensor with two parts. The implanted part consists of a capacitive absolute pressure sensor (ASIC) and its telemetric electronics (Figure 2-8). The telemetry has a microprocessor as the main control unit of this part. The second part is the external reader station which is controlled with a PC. The system can measure both blood pressure and temperature at the same time [5].
All the different parts, including pressure sensor, implant electronic and external reader station are commercially available which make the whole system very cost-effective. The pressure sensor is catheter based and easy-to-implant. It has an integrated signal conditioning unit and an analogue/digital converter.

First step in the sensor packaging is covering the whole implantable part by an UV-glue. This will fix all components and protect them in the further process of encapsulation. Then a layer of two-compound epoxy resin is added. Epoxy resins are well known for being suitable for electronic implant encapsulation [45]. When this stage is completed, three links are mounted on the system with 120° spacing (see Figure 2-9). These are for achieving a monolithic encapsulation. The encapsulation is completely biocompatible. It has 28mm diameter and 4.5mm height.

The telemetry part is micro-welded to the sensor via a four-line micro cable. The micro cables are mechanically protected by means of a polyurethane (PU) medical catheter.
tube. The resulting unit is stiff enough to be pushed into the blood vessel. Having 120mm length makes the sensor easy to handle during the implantation process. Considering some handling issues related to the implantation process, the length of the connecting micro cable was selected to be 120mm. The sensor tip is covered with medical grade silicone. The outer side layers of the whole system are coated by Parylene-C thin film approximately 3µm thick. The sensor tip is haemocompatible due to biocompatibility of Parylene-C. The film also reduces friction, so the sensor can be pushed easier into the catheter during the implantation process.

There are two different evaluated designs. In the first one catheter tube is mounted in front of the pull relief (Figure 2-10) while in the other one the sensor is cased with the tube up to the membranes at its tip (Figure 2-11).

![Figure 2-10](image1.png) [5]: The tube does not cover the sensor

![Figure 2-11](image2.png) [5]: The sensor is covered by the tube up to its membrane

These two configurations operate similarly but the system in approach b (Figure 2-11) in spite of 0.2mm bigger diameter can be handled easier and better protect the sensor. The sensor is designed for implantation in femoral artery and has been tested both in vivo and in vitro. In vivo measurements in eight sheep have been successfully reported [46].

### 2.2.2 Intraocular pressure sensor

Glaucoma is one of the most common eye disorders [6]. In this disease the patient suffers from loss of eyesight due to a damaged optical nerve [6, 11]. One of the risk factors of this disease is high intraocular pressure (IOP) caused by accumulation of ocular fluid [3, 6, 11]. So long-term IOP fluctuations’ monitoring can be very useful for these
patients. Unfortunately common clinical techniques cannot record IOP values faithfully and monitor its fluctuations properly. This limits the evaluation of glaucoma progression. Readouts from current tonometers including contact and non-contact are not precise enough because they are strongly dependent on the size and mechanical properties of different eyes [47, 48].

In 1990 a small pressure sensor assembled in an artificial intraocular lens was introduced [49]. Another commercially available IOP sensor was introduced in 1998. It consisted of a silicon lens with an IC, a coil and a pressure sensor [50, 51]. A moving silicon membrane changes a capacitance; this varies a resonant frequency by means of the micro coil in an LC-resonant circuit [3].

In 2006 a sensing technology has been proposed by Po-Jui Chen et al [52]. This system is all-mechanical and un-powered. The sensor has no electrical circuitry or external powered signal transduction. This acts as the sensing element of pressure sensor. The system has an implantable MEMS pressure sensor for IOP monitoring. This sensor has a curved hollow tube which is standing freely, (Bourdon tube) [53]. It is a Parylene based tube which is deformable. The pressure difference between the inside and outside of the tube/channel causes a deformation which is visible at its free end with a pointing tip. The movement of this tip has a linear relation with the pressure difference. By setting the pressure inside the tube to a predetermined constant, the outside pressure can be measured [6]. The visible deformation of the implanted tube makes it possible to measure intraocular pressure directly through optical observation. The sensor can precisely detect variation of IOP due to glaucoma, diurnal rhythm and atmospheric pressure change. By utilizing various geometric designs amplification of tube deformation and modification of its in-plane deformation trajectories would be possible (see Figure 2-12 [6]). These different geometric designs can increase sensitivity as well as optimizing device dimensions for clinical use. In addition, such symmetric designs can be more accurate due to their minimized out-of-plane deformation.
Possible layouts for designing IOP sensor. Design (b) is good for increasing the angular rotation at the pointing end. Design (c) with its long arm increases the lateral movement.

This pressure sensor is designed to be implanted on the iris (see Figure 2-13 [6]). Visibility of the deformation of implanted tube and transparency of cornea makes it possible to measure IOP directly through optical observation. Proper placement of the sensor makes it very sensitive to fluctuations of IOP.

Furthermore, the sensor packaging should necessarily ensure that it works properly after implantation. By the help of micromachining technology a fully packaged device is fabricated on a monolithic substrate which consists of the following components: One sensor platform providing support to the IOP sensor; the others are the bottom posts with anchoring feet to fix the implant on the iris and a perforated flexible Parylene membrane to avoid damages during implantation (See Figure 2-14).
In order to have a flexible tube with thin walls Parylene-C is used as the structural material for the tube. It is sufficiently flexible, chemically inert and biocompatible [54, 55]. Parylene film is deposited by CVD (chemical vapor deposition) technique at room temperature.

Two such types of IOP sensors have been microfabricated on monolithic substrates. Both of these two types are of high-sensitivity. One is short handed and the other is a long handed variant. The former is a simplified version which just has the Parylene tube (see Figure 2-15 left). The latter with a smaller form factor has all the necessary components for implantation were shown in Figure 2-15 right.

The IOP sensors were tested in vitro. A reference pressure of 1atm is applied while the pressure on the other side is changing the pressure surrounding the sensor. Then the pressure is measured upon the structural deformation and is compared with numerical readouts from a pressure gauge. The measurements are done in both air and liquid media. A linear pressure response is reported for both high-sensitivity sensors. The needle-

---

1 See Chapter 4, section 4.4 (VDP Process)
implantable sensor with long-armed-tube had a relatively linear pressure response. The sensor also has been tested in vivo. It is implanted in a rabbit eye model in a five-minute long process. After one month the device had not been moved. In addition ex vivo tests by both spiral-tube and long-armed-tube sensors had been successful in registration of IOP variation in the intraocular environment. Furthermore, sensors show good repeatability in pressure cycles.

2.2.3 Bladder Pressure Sensor

The most useful way of assessing the bladder condition is to measure its pressure [16]. Most of the usual bladder measurements are done by means of catheterization in the physicians’ offices. This technique is non-physiological and also painful. So work has been done on ambulatory urodynamics to achieve a more physiologic pressure monitoring. In ambulatory urodynamics monitoring the patient can move while the bladder is allowed to fill naturally during test procedure [66].

There had been some works on telemetric systems for measuring bladder pressure. Those systems were not favorably used, because of the size and weight of the implant package. Furthermore, the system implantation and removal needed surgery which made the method more inconvenient [16].

Mainly, pressure sensors used for measuring bladder pressure have a catheter and the sensor is placed at its tip [11]. This method is not favorable because of the wires connected to the sensor. These sensor wires need continuous care to maintain sanitation and to avoid entanglement during long period of time.

Ekapon Siwapornsathain et al reported a system with a piezoresistive pressure sensor and a small RF transmitter inside a catheter [16]. This pressure sensor has a silicon nitride membrane with integrated piezoresistors. An oscillator is used to transmit the pressure information. The frequency of oscillation is determined according to the pressure sensed by the piezoresistors. Power supply of the system is a tubular 3V lithium battery. The device produces an RF carrier frequency which can be varied in a specific range by a modulator. An external radio can demodulate the transmitted frequency for electronic or audio recording.
The system has been tested \textit{in vitro}. The sensor is put in a pressure chamber and operated while measurements are performed by a multimeter. The receiver has an amplifier and an audio output. The audio tone is analyzed to determine the pressure.

For the packing of the system, the pressure sensor and IC are placed on a printed circuit board and wire-bonded (see Figure 2-16). The whole system is coated with a film of medical grade silicone to make it biocompatible.

![Figure 2-16](image)

\textbf{Figure 2-16 [16]}: Schematic of the device which is coated with a thin layer of silicon: The electronics are placed on the PC board connected to a 3V lithium battery. The latex balloon is assembled to the system to prevent the unwanted expulsion of sensor out of the bladder.

Buoyancy capacity of the device was also tested by submerging it into saline. Having greater density than water, the device sinks to the bottom. To overcome this, a balloon is attached to the end of the device and inflated by means of a syringe pump. The device has been tested inside the bladder of a female patient. Implantation process had been successful and the external receiver station could have detected the transmitted data.

Song \textit{et al} have presented a new physiological method for abdominal pressure measurement [66]. This is also an ambulatory urodynamics monitoring (AUM). The system consists of the following parts: a disposable double-lumen catheter, a pressure transducer, a recorder for signal conditioning, hardware to save the measured data and software to monitor and analyze the physiological parameters. The system measures the bladder and abdominal pressures by means of two catheter-based transducers and at the same time measures surface electromyograph signals (EMG). These measurements are performed during daily activities. The catheters are connected to electronics which consists of a signal conditioning circuitry and flash memory. The signal conditioning unit has an instrumentation amplifier and a filter to reject noise (analog part in Figure 2-17). It pre-processes the measured pressure and EMG signals. The signals from this unit are then
sent to an analog-to-digital converter to be digitized and then transferred to flash memory for storage [66].

**Figure 2-17** [66]: System block diagram for measuring bladder pressure and surface EMG signals

### 2.2.4 Pressure sensor of Abdominal Aortic Aneurysms

The largest artery in the abdominal cavity is the abdominal aorta [13]. Sometimes due to a disease or weakened vessel wall, blood pressure may cause a balloon-like bulge in abdominal aorta; such a disorder is called an Abdominal Aortic Aneurysm (AAA) [11]. It can cause death if the bulge in the vessel burst. Burst of bulge is most probable in large aneurysms [13]. AAA can be treated by implantation of a foldable prosthesis (stent) in aorta through an inguinal incision (see Figure 2-18) [3].
This operation mainly aims to lower the pressure in the aneurysm sac and to avoid rupture [30]. Unfortunately due to incomplete exclusion of the sac or leakage, the sac will be pressurized continuously [3, 11, 30]. Such failures can be detected and therefore treated by means of a pressure sensor to monitor the pressure inside the abdominal aorta. Different groups have been working on such sensors.

Schlierf and colleagues presented a system that consists of an implantable capsule and external readout station [31]. The capacitive pressure sensor, microchip and a ferrite coil are integrated inside the capsule and transmit the measured data to a readout station wirelessly [31] as shown in Figure 2-19 [31].

**Figure 2-18** [3]: placement of a stent in abdominal aorta for treatment of its aneurysm

**Figure 2-19[31]**: sensor system: pressure sensor, microchip and LC resonant circuit are in the implantable capsule (left) and another LC resonant circuit and electronics are in the readout station (right) to receive data sent by sensor
The implant system does not need an internal power supply, because the required energy is induced to the coil at frequency of 133 KHz. The implanted capsule is able to preprocess the measured data before transmitting them to external station. The data are converted to a digital data stream and then transmitted to the readout station. This minimizes the errors during the transmission process or from interferences between the sensor and the external unit. The implanted unit has an integrated temperature sensor. This makes it possible to note the temperature variations and correct the possible errors numerically.

A silicon layer and a Parylene-C layer are covering the capsule to make it biocompatible and stiff enough for using the capsule as an implant [31]. Figures 2-20 and 2-21 show the inside of the capsule and the encapsulation respectively.

**Figure 2-20 [3]:** inside of capsule: from left to right: Microchip, Capacitor and Ferrite coil

**Figure 2-21 [3]:** implantable sensor capsule consists of two-layer-encapsulation: Silicon and Parylene-C

Another implantable pressure sensor for use after endovascular aneurysm repair (EVAR) is presented by Ohki et al. The externally powered pressure sensor is an intraoperative EndoSure sensor with pulse pressure measurements. In this sensor the electronic components are surrounded by a nitinol basket (see Figure 2-22). The basket has no electrical functionality but is only used to keep the sensor in the center of the sac.
According to the results of *in vivo* tests the sensor is capable of working several years with an extreme stability. The sensor can be calibrated during its deployment in the aneurismal sac and does not need recalibration. It consists of flexible plates which encapsulate inductor windings in a hermetically sealed reference cavity (See Figure 2-23). Any variation in the ambient pressure of the sensor will displace the plates, resulting in a change of the capacitance and the resonant frequency of the sensor. This variation in resonant frequency can be detected by an external receiver which can be used to measure the pressure in an aneurysm sac [30].

---

**2.2.5 Intracranial Pressure Sensor**

Elevated brain pressure is one of the symptoms observed in some brain diseases [56]. Human intracranial pressure (ICP) is normally between -5 to 10torr\(^1\) relative to the ambient atmospheric pressure [34]. When the head is at the same elevation as the heart, the pressure is usually at the positive side. Devices which are currently used in clinical applications to monitor intracranial pressure are not telemetric. They all need a connection between the head and the external environment. Such systems have some limitation in duration of monitoring, and suffer from high risk of infection. Different

---

\(^1\) 1torr=1/760 atm ≈133.32 Pa
groups are working on implantable telemetric systems which are capable of long-term ICP monitoring and wireless data transmission with higher reliability and lower risk of infection [35]. Although the piezoresistive pressure sensors are favorable for disposable catheters use, they have high power consumption which is a drawback for an implant. In addition, these sensors react on temperature changes [17]. At present very small capacitive sensors with an integrated temperature sensor on the chip can be fabricated. B. Flick and co-workers made a portable intracranial pressure (ICP) sensor with such capacitive sensors. The system is useful for patients in the neurosurgical field. It consists of an implant unit which measures intracranial pressure and temperature, a portable recorder and an external data processing station. The data recorder has the same sensor to measure atmospheric pressure and outside temperature because of dependency of ICP to atmospheric conditions. It displays each single value of the measured signals. Data processing and monitoring is done afterwards in the external station. Figure 2-24 shows the block diagram of the whole system. The recorded data consist of the brain and atmospheric pressure and the temperature. It can be displayed on an LCD monitor and be saved on data storage equipment. The equipment can be easily changed by the patient and the recorded data can be transferred to a computer afterwards for further analysis. Another sensor system is also available with a stainless steel or plastic package. This pressure sensor for intracranial pressure measurements has a flat droplet form as can be seen in Figure 2-25 [23]. It does not damage the ambient tissue due to its shape. It has a weight of only 3g height less than 2.5 mm, and can measure pressure with an accuracy of between 0 to 4 Pa in different sections of the brain.
Figure 2-24 [64]: Block diagram of sensor system: (a) intracranial Unit, (b) Portable Unit and (c) is the stationary unit.

Figure 2-25 [23]: The intracranial Pressure sensor with less than 2.5 mm height and 3 g weight
A piezoresistive pressure sensor for ICP monitoring is made by Codman. This American company works specially on diagnostic and therapeutic products. The sensor membrane is connected to the electronic interface through a nylon wire. E. Morganti *et al.* clarified that the nylon wire prevents the system from being fully implanted [26]. By means of a telemetry wireless system this problem can be solved. The telemetry utilizes a resonant LC circuit. The sensor measures the pressure by inductive coupling of its resonance frequency and that of an external coil [26].

M.R. Tofighi *et al.* have reported a non-invasive microwave intracranial pressure sensor with a mobile microwave monitor [34]. The sensor has a MEMS capacitor and an oscillator. Any variation in intracranial pressure results in a change in its oscillation frequency. The sensor output is coupled to a Bluetooth chip antenna and can be displayed by an external unit. The microwave oscillator has been selected due to several reasons such as high sensitivity, possibility of radiation by a small antenna without absorption in the implanted package and the skin. The structure is placed in cylinder shaped titanium cage to ensure biocompatibility and magnetic resonance imaging (MRI) compatibility. The system is driven by a 3V lithium battery that must be placed under the skin during the implantation process (see Figure 2-26).

![Figure 2-26](image)

**Figure 2-26 [34]:** The sensor is placed on the brain by an implantation surgery to measure the intracranial pressure.

Tofighi and co-workers have later introduced another implantable pressure sensor which can measure ICP non-invasively after neurosurgery [24]. The piezoresistive pressure sensor transmits data on a wireless link through the scalp. The device has a stainless steel case with electronics inside. It is to be implanted in a burr hole in the skull and is fastened to it with screws. The sensor is placed on the dura mater (see Figure 2-27) [24].
electronics inside the case consist of two printed circuit boards in cylindrical structure, and side legs for electrical connection as well as for stability reasons. Ground, voltage supply and pressure signal are transmitted by these connections. To test the performance of the sensor in similar conditions as the device embedded in a highly absorptive tissue medium, the device under test is coated with medical grade silicone sealant, covered by a gel, and submerged in water.

**Figure 2-27 [24]:** (a) shows the device without case (b) is with the stainless steel case included
3 Medical Background

The present master thesis is part of a sensor project that investigates the use of implantable pressure sensors for use in Hydrocephalus. In the head there are empty spaces around and inside the brain filled with cerebrospinal fluid. These empty spaces include the subarchonoid spaces (see Figure 3-1) and the ventricle system (Figure 3-2) [9, 13]. Cerebrospinal fluid (CSF) is a clear liquid [13, 21, 26] with blood-plasma-like composition [21, 26] and density of 1005-1009 g/l [26]. It feeds the brain as well as cleaning and protecting it [9]. It is normally produced and absorbed with rate of 500 ml each day [9, 26].

Figure 3-1 [13]: Meninges have three membranes: Dura mater, Archnoid and Pia Mater. The subarchonoid cavity is the empty place between arachnoid mater and pia mater.

Figure 3-2 [29]: Human brain ventricular system

1 “Water on the brain” [8, 29]
3.1 What is Hydrocephalus

Any disturbance in the rate of production and absorption of CSF can result in a hydrocephalus condition [9, 20] (Figure 3-3). The liquid pressure in the brain should normally be between 0.5KPa to 1KPa [83, 84] (see Table 3-1). Symptoms of increased pressure are age-dependent but can mainly be summarized as sunsetting eyes [8], larger than normal head in newborns, and vomiting, nausea, imbalance in walking [8, 9]. Bleeding inside a brain ventricle, a head injury, brain infection or a brain tumor can cause hydrocephalus. In addition about 0.2% of live births [8, 9] give rise to a hydrocephalic child.

<table>
<thead>
<tr>
<th>Condition</th>
<th>Pressure (KPa)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Extra-Low Pressure</td>
<td>0 - 0.1</td>
</tr>
<tr>
<td>Low-Pressure</td>
<td>0.1 - 0.5</td>
</tr>
<tr>
<td>Medium-Pressure</td>
<td>0.5 - 1</td>
</tr>
<tr>
<td>High-Pressure</td>
<td>1 - 2</td>
</tr>
</tbody>
</table>

Table 3-1 [83]: shunt valves pressure. Most of shunt valves used for Hydrocephalus treatment have a self-regulating valve. They can adjust themselves so that suitable amounts of CSF will be drained due to the intracranial pressure.

3.2 Hydrocephalus Treatment

Surgical installation of a tubing system called a shunt, is the common treatment for hydrocephalus to drain the excess CSF [8, 9, 18, 19, 21, 26]. The shunt takes the CSF from the ventricle to another part of body [8, 9, 18] such as the abdominal cavity, jugular
vein or the bladder (see Figure 3-4) [18]. The shunt has a valve which allows the liquid to flow only in one direction [8, 26].

Unfortunately, some failures may occur in the shunt systems, such as disconnection of tube, over-drain or under-drain, or blockage of the tube [9, 18, 19, 26]. Work is proceeding to improve shunt system through different approaches: MEMS applications such as programmable valves, smart shunts with the help of flowmeters [18], and shunts integrated with magnetic microactuators to reopen clogged tubes [19]. A system which can be installed permanently deep in the brain for measuring brain ICP would also be beneficial. The drawback of this system is that it can not respond immediately and in an effective manner to the changes in intracranial pressure. This can be solved by using a closed loop system comprised by an intracranial pressure sensor, a control system and a micro-pump that can move CSF based on the data collected by the sensor [26].
4 Coating Properties and Challenges

It is important to have an appropriate package to obtain high reliability [25] for long duration. The application of a pressure sensor determines what kind of packaging is useful [25]. There are special challenges for packaging of an implantable microsystem. It should be biocompatible, small enough and able to protect the implant for sufficient time in vivo [27]. In presence of body fluid it should be non-reactive and biostable. It should not chemically react to substrate and contaminate it with by-products or outgassing. The cure temperature of the coating should be compatible with the coated item. The coating should not have any negative effect or make any distortion to the surface. It should be able to withstand the sterilization process, and in addition to all above, it should not change the mechanical characteristics of the substrate significantly. Sufficient robustness and adherence is required to avoid undesired detachment of particles from the coating or the coated surface. A pressure sensor coating must not mechanically load the active membrane or interfere with the device functionality. In addition it should protect the delicate elements from moisture and other external undesired factors [68].

The following properties are targeted for the coatings:

- Insulating material
- Biocompatible
- No stress
- Chemically inert
- Adheres perfectly to substrate
- Has coefficient of thermal expansion close to Si
- Resistant against wear and scratches.

Three coatings $\text{TiO}_2$, $\text{Al}_2\text{O}_3$ and $\text{Parylene-C}$ were selected. They were deposited in two different thicknesses for each. $\text{TiO}_2$ and $\text{Al}_2\text{O}_3$ were deposited by ALD (atomic layer deposition) in low temperature ($150^\circ\text{C}$) by the chemistry department at University of Oslo. The $\text{Parylene-C}$ coating has been done by SCS (Specialty Coating Systems™).
4.1 Properties of Al₂O₃ and TiO₂

Titanium and titanium oxide are known as very good biocompatible materials [59, 79]. Dental and orthopedic implants need to be mechanically strong, chemically stable and also biocompatible [60, 61, 85]. Satisfying these requirements, pure titanium and its alloys are commonly used for such implants. These properties are caused by spontaneous formation of a thin oxide layer on titanium surfaces [36]. This adherent layer passivates the material. Titanium dioxide, Titania, is known as an antimicrobial coating. Titania thin coatings exhibit self-cleaning and disinfection properties when exposed to UV radiations. This stems from the photocatalytic activity of Titania. Having such characteristics, Titania is favored in various applications such as medical devices, food preparation and sanitary-ware surfaces and air conditioning filters [80].

Aluminum Oxide, Al₂O₃, is one of the most commonly used engineering ceramics. Alumina has excellent properties such as wear resistance, chemically resistance against alkali and acid attacks and reasonable fabrication price [75]. In medical application it is known as a biocompatible coating with high corrosion resistance [65, 78]. Al₂O₃ with high purity is used in implants especially in hip replacement surgeries [33] (p. 555), [76]. There are some other medical applications of Alumina such as bone spacers and tooth replacements. When large part of bone has been removed, it can be replaced by rings made of porous alumina. These rings are concentric around a metal pin which is inserted inside the remaining part of the bone. The newly formed bone can grow into the porous implant [78].

Al₂O₃, SiO₂ and TiO₂ are among few ceramic oxides applicable for bioinert use [37]. Bioceramics have most of the desired properties of an excellent implantable. They are not allergic, toxic, inflammatory or carcinogenic [33] (p.553). Al₂O₃ is the most used inert bioceramic now and it has good mechanical behavior.

In comparison to Al₂O₃, TiO₂ is more biocompatible in some applications that involve blood contact, but has worse mechanical properties. To overcome this drawback it is possible to reinforce it by means of Al₂O₃ platelets without any change in bioinert characteristics [37].
4.2 Properties of Parylene-C

Synthetic polymeric materials are superior to other biomaterials in different aspects such as easy manufacturing process, possibility of secondary process, reasonable production cost and favorable mechanical and physical properties. Considering these characteristics polymeric materials are widely used in medical world. Disposable medical supplies, polymeric prosthetic materials, dental materials and implants are some examples of their applications [33] (p. 581).

Parylene is a member of poly-para-xylylene family [13, 70]. The basic member of this family is Parylene-N which is a completely linear and highly crystalline material. If one of the aromatic hydrogens in Parylene-N is substituted with a chlorine atom, Parylene-C is formed (Figure 4-1) [70]. Parylene-C is a colorless material with good electrical and physical properties as well as low moisture and corrosive gas permeability. Combination of all above makes it a good candidate for medical surface modification applications [68]. Some properties of Parylene-C are discussed below in more details.

Figure 4-1: (a) Parylene-N (b) Parylene-C

The water vapor transmission rate for Parylene-C is superior to the most common polymeric materials like Acrylic, Epoxy or Polyurethane. Circuit boards are protected against corrosion or salt deposition when coated by Parylene-C. Uniform and conformal Parylenes’ coatings are able to completely cover the electronic packages and circuit boards [70]. These insulation coatings are transparent and evenly distributed on surfaces [69].

Parylenes are insolvable in all organic solvents up to 150°C [70]. Parylene films are also chemically resistant to inorganic reagents and acids attacks [68]. Parylene-C at 175°C is dissolvable in chloro-naphthalene but it resists permeation of most solvents [70]. Parylene coatings are used for coating of implantable and non-implantable medical
devices. They protect devices from biofluids, biogases and moisture as well as having biocompatible surface for the contact tissue. Parylene-C can withstand different sterilization processes such as steam autoclave, gamma and e-beam irradiation, hydrogen peroxide plasma, and ethylene oxide [70].

Having low mass Parylene films have minimal damping and loading effects on substrates [68]. Parylene-C has density of 1.289g/cm$^3$. It is very flexible (Young’s modulus is approximately 2.76GPa) [70].

Parylene-C applied on a metallic substrate does not adhere well enough to be used in an implant. The adherence can be increased by means of multilayer silicon dioxide, plasma-polymerized methane and Parylene-C [73].

4.3 Atomic Layer Deposition Process

Atomic Layer Deposition (ALD) is a chemical process in gas phase used to make thin coatings with accuracy of 1nanometer [13, 67]. This technique is based on sequential self-terminating gas-solid reactions and used for manufacturing inorganic layers [67]. ALD is unique among film deposition technique in the sense that it is capable of coating extremely complex shapes with the conformal material layer of high quality. A conformal coating is the one which covers all sides of the part to be coated and penetrate into its holes and tubes as well as covering sharp edges and points [69]. ALD is attracting more attention mainly due to its application in microelectronic devices [72].

In ALD technique material layers are deposited on surfaces by iteratively performing the following steps. First reactant A is introduced to the reactor to form a material monolayer on the substrate. This is a self-terminating reaction. Then by-products and non-reacted reactants are purged. As the third step reactant B will be introduced or another treatment will be performed to re-activate the surface for the reaction of the first reactant. This can be thermal treatment for instance. Forth step is purge or evacuation. These steps are known as ALD reaction cycle (Figure 4-2) [67].
As an example for deposition of a metal oxide from a metal iodide and oxygen, the ALD cycle consists of the following sequences.

1. The metal pulse (a molecular monolayer of metal is adsorbed to the surface);
2. Purging pulse (non-reactive gases are wiped out);
3. Oxygen Pulse (Oxygen gas reacts with the previously adsorbed metal and forms a new monolayer);
4. Purging pulse.
Chemical reactions in ALD do not mix [71]. Repeated monolayer depositions will therefore result in an extremely conformal and uniform film thickness. The use of self-terminating processes causes temperature and other parameters to have minimal effects on the ALD process [67].

Growth per cycle (GPC) determines the amount of material which adds to the surface in each ALD reaction cycle. To reach the desired amount of deposition the reaction cycles will be repeated.

Al₂O₃ is one the most frequently used ALD-grown materials [62]. Alumina deposition can be done by different ALD procedures with relatively short cycle times [63].

4.4 Vapor Deposition Polymerization Process

Parylene coating is done by a process called vapor deposition polymerization (VDP) [68, 69, 70]. In this process a dimer of Parylene called di-para-Xylylene (DPX) is vaporized at 1.0torr in 150°C. Then the gas is heated up to 680°C to cleave molecularly¹ or pyrolyze² it. This results in a reactive monomer gas (PX). This cleavage takes place in a pressure of 0.5torr. Then the gas is pumped into the deposition chamber at room temperature with 0.1torr pressure. Xylylene spontaneously polymerizes (PPX) on the substrate to be coated. It forms a conformal colorless film on the substrate without heating it more than a few degrees above the environment temperature. The deposition process is driven by the pressure gradient in three different stages [68, 69, 70]. Figure 4-3 shows the VDP process schematically.

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¹ *In chemistry splitting (a complex molecule) into simpler molecules is called cleave* (American heritage dictionary)

² *Decomposition or transformation of a compound by heat called pyrolysis.* (American heritage dictionary)
Since the film is formed at room temperature, there is no difference between cure and room temperature and therefore the surface would not be exposed to any stress [68]. The amount of vaporized DPX and the time it resides in the vacuum chamber determine the coating’s thickness. This leads the process to be an accurately controlled one. For film formation between 0.1 to 75 microns one single coating cycle is sufficient [68].

### 4.5 Biocompatibility

When talking about usage of new materials in medicine the first issue that comes to one's mind is its biocompatibility [65]. Biocompatibility is the acceptance of a material by a living tissue or the body as a whole [2, 33]. It can be an issue for an implant, or any medical device such as surgical instruments, dental implants and instruments and any other medical implants and prosthesis. How long and where the medical device can be utilized is determined by its biocompatibility. A long term implant device is assumed to work with proper performance and acceptable functionality in the host without causing any adverse effect (locally or systematically) [2]. Materials used in microchips are basically bioincompatible. An ion concentration corresponding to a 0.9% saline solution, will seriously influence a pressure sensor where the piezoresistors are in the kilo-ohms range. The pressure sensor microchip therefore needs a biocompatible coating all over its surface in order to work properly inside the body. However it should be emphasized that
the coating procedure should not degrade the sensor functionality and biocompatibility. Some important properties of an ideal coating are edge covering, peel resistance and integrity of the regular film. Adsorption of proteins and adhesion of cells onto polymeric coatings are affected by surface properties [22]. Many factors can affect biocompatibility such as corrosion resistance in body fluid which is one of the most important factors [65].

4.6 Biofouling

Different mechanisms such as protein adsorption, cellular adhesion, platelet adhesion/activation and surface-induced thrombosis can cause biofouling in an implanted device; depending on the host’s environment [2, 10]. Some failures may happen due to biofouling [32]. It is frequently reported that the sensitivity of sensors decrease when placed in a tissue [10]. These failures can be categorized into two groups of component-based and biocompatibility-based failures. Failures like lead disconnection, short circuits and membrane delamination are examples of the former. And membrane biofouling, electrode passivation, fibrous encapsulation, and membrane biodegradation are examples of the latter. Although there is no study which can clearly give us a convincing ranking about the effectiveness of the above mentioned issues on the in vivo sensor failure, the membrane biofouling is certainly known to be destructive to the sensor functionality [32].
5 Sensor System

Sensors are usually defined as devices that respond to a received stimulus [7] (p.1). To give a more specific definition, they can be defined as energy/signal converters which convert energy/signal from one form to electric form [7] (p.2), [33] (p. 725).

The biomedical sensors usually have an input signal which represents a biomedical variable and as output they usually generate an electrical signal. Therefore the biomedical sensor can be regarded as an interface between a biological and an electronic system which should not degrade the performance of these two systems. Both electronic and biological systems play an important role in the performance of a biomedical sensor [33] (p. 725).

5.1 Piezoresistive Pressure Sensor

For making a pressure sensor a sensing element is needed, usually an elastic diaphragm, which is deformed due to applied differential pressure, and a detector which responds to this deformation. In the most common [11] type of MEMS pressure sensors, an elastic silicon membrane is normally fabricated by bulk or surface micromachining [1, 11]. The sensors used in this thesis have been made with bulk micromachining. Piezoresistive resistors are placed in the membrane, and coupled as part of a Wheatstone bridge configuration [7] (p.344). The resistors should be in the areas where the membrane is exposed to the maximum stress or strain. Their resistivity is changed by variation of strain and stress on the membrane. This change is measured by corresponding voltage change in the output of the bridge.

5.1.1 Piezoresistivity

The electrical resistivity of a material depends on its atomic positions and motions. Strain can change the resistivity by changing the atoms’ positions. The dependence of electrical resistivity on strain is called piezoresistivity [28] (p. 470). Atomic constituents and their positions determine the electronic states of a material. Unlike free atoms, in crystalline materials these states are in form of bands (energy bands). In other words free
atoms have some discrete energy levels while crystalline materials have quasi-continuous available energies [28, 74]. The difference between insulator, semiconductor and conductor materials can be explained by their different available energies for their electrons. The electrons in the valence band of an insulator are separated from the conduction band by a large gap while in conductors there is an overlap between the conduction band and the valence band. In semiconductors a small gap exists between the valence and the conduction band. In this case an excitation can provide enough energy for electrons to bridge the gap [74].

The Pauli Exclusion Principle says that for any two electrons the set of quantum numbers cannot be the same. This is correct not only for electrons but also for all fermions. Fermions are particles whose spin quantum number is a positive half integer value; unlike that of bosons which is zero or positive integer. Electrons, protons and neutrons are fermions. Photons are examples of bosons with spin value of $s=1$ [82]. The electrons in crystalline materials fill the energy band according to Pauli Exclusion Principle from lowest to highest filled energy level. In metals the highest filled energy level occurs in the middle of a band. So close to the highest filled energy state there are a large number of empty spaces. When an electric field is applied, the occupancy of these levels shifts slightly. This movement of charge carriers is called electric current [28] (p.470).

When a stress is applied to the metal, atomic positions are changed. This results in slight distortion in energy bands and therefore small changes in the amount of conduction. This is called piezoresistive effect [28] (p.470).

Semiconductors in a thermal equilibrium have both positive and negative carriers. The highest energy level called valence band is filled with valence electrons. The higher energy quantum-states which electrons can reach are called conduction band and in between there is a band gap without any allowed quantum state.

The electrons occupying the conduction band are called conduction electrons with a charge of $-q_e=-1.6\times10^{-19}\,C$. The empty state near the top of the valence band is called hole which is considered a positively charged carrier with a charge of $+q_e=+1.6\times10^{-19}\,C$ [28] (pp. 353, 354). When applied stress changes the atomic positions, the energies move slightly. It can have a huge effect on conductivity properties. So the conductivity may
increase in a direction and decrease in another. This is the principle of constructing a piezoresistance strain gage. For a long resistor the relation between piezoresistivity and changing in resistance can be formulated as following equation:

\[
\frac{\Delta R}{R} = \pi_l \sigma_l + \pi_t \sigma_t
\]

Where \( R \) is the resistance, \( \pi \) is the piezoresistive coefficient and \( \sigma \) is the stress. Subscripts \( l \) and \( t \) refer to longitudinal and transverse stresses respectively [28] (p.473).

5.2 Bending of a pressure membrane

Any pressure applied across the membrane can cause bending. Transverse Poisson contraction is inhibited if the membrane (plate) is wide enough and well supported (Figure 5-1) [28] (p. 219).

![Figure 5-1: (left) clamped plate (right) bending of a clamped plate by applying pressure](image)

There is a plane called neutral plate which is neither compressed nor stretched. Stress and strain are proportional to the \( z \)-coordinate. If the \( z_c \) is the \( z \)-coordinate of neutral plane, equation 5-2[28] (p.212) shows the stress in the bent plate.

\[
\sigma_x = -\frac{(z - z_c)E}{\rho}
\]

where \( \sigma_x \) is the stress in \( x \)-direction, \( \rho \) is the radius of curvature and \( E \) is the Young modulus.

A square plate subjected to an applied pressure has the maximum deflection at the center of the plate. Equation 5-3 shows the displacement at the center of the plate, \( z_0 \).
Equation 5-3: \( z_0 = \frac{(1-v^2)}{E} \frac{6L^4}{\pi^2 t^4} P \)

where \( v \) is the Poisson ratio, \( t \) is the plate’s thickness and \( P \) is the applied pressure. By replacing \( D \), flexural rigidity of the plate [28] (p.212), (equation 5-4) in equation 5-3, equation 5-5 will be obtained.

Equation 5-4: \( D = \frac{1}{12} \frac{E}{(1-v^2)} t^3 \)

Equation 5-5: \( z_0 = \frac{1}{2D} \frac{L^4}{\pi^2} P \)

This shows the relation between displacement at the center of the plate, applied pressure and stiffness.

5.3 Coating Effects on Membrane Sensitivity

Sensitivity of the sensor can be changed due to movement of neutral plane and changes in stiffness after coating. Sensitivity of a piezoresistive pressure sensor is proportional to piezoresistive coefficient of the sensor, stiffness of its membrane and neutral plane coordinate. When piezoresistors have a bridge configuration the output voltage is equal to the effective piezoresistive coefficient times the average stress [87] (equation 5-6).

Equation 5-6: \( \frac{\Delta V}{V} = \pi_{\text{eff}} \sigma_{\text{avg}} \)

While average stress is proportional to inverse of \( D \) times \((z_{PZR} - z_c)\), where \( z_{PZR} \) and \( z_c \) are the z-coordinate of piezoresistors and neutral plane respectively. Change in sensitivity can be obtained by equation 5-7.

Equation 5-7: \( \frac{\Delta s}{s} = -\frac{\Delta D}{D} \frac{\Delta z_c}{z_c} \)

The neutral plane has zero stress. In multilayer plates the z-coordinate of the neutral plane changes (See Figure 5-2 for coordinate system). In an n-layer clamped plane, when the pressure is applied in z direction; the z coordinate of the neutral plane can be obtained by Equation 5-8 [88].
**Figure 5-2:** A clamped plate. The z axis is normal to the plate.

**Equation 5-8:**

\[
z_c = \frac{E_1}{1-V_1^2} t_1 \left( \frac{t_1}{2} + \frac{E_2}{1-V_2^2} t_2 \left( \frac{t_2}{2} + \frac{E_3}{1-V_3^2} t_3 \left( \frac{t_3}{2} + \ldots + \frac{E_n}{1-V_n^2} t_n \left( \frac{t_n}{2} + \frac{z}{2} \right) \right) + \ldots + \frac{E_n}{1-V_n^2} t_n \right) \right)
\]

By having \( z_c \), the moment of inertia can be calculated from Equation 5-9 [88].

**Equation 5-9:**

\[
D_i = \frac{E_1}{1-V_1^2} \left( \frac{t_1}{12} + t_1 \left( \frac{t_1}{2} - z \right) \right) + \frac{E_2}{1-V_2^2} \left( \frac{t_2}{12} + t_2 \left( \frac{t_2}{2} - z \right) \right) + \ldots + \frac{E_n}{1-V_n^2} \left( \frac{t_n}{12} + t_n \left( \frac{t_n}{2} - z \right) \right)
\]

Changes in sensitivity due to coating of pressure sensors are calculated by these two approaches: movement of the neutral plane and changes in bending stiffness.

### 5.3.1 Neutral Plane Movement Due to Coating

The effect of changing the neutral plane coordinate is discussed below. In Table 5-1 mechanical properties of different coatings and silicon are listed.

<table>
<thead>
<tr>
<th>Material</th>
<th>Silicon</th>
<th>TiO₂</th>
<th>Al₂O₃</th>
<th>Parylene-C</th>
</tr>
</thead>
<tbody>
<tr>
<td>E(GPa)</td>
<td>160[58]</td>
<td>230 [80]</td>
<td>400[77]</td>
<td>2.75 [70]</td>
</tr>
<tr>
<td>ν</td>
<td>0.23[58]</td>
<td>0.27 [80]</td>
<td>0.26 [77]</td>
<td>0.4 [86]</td>
</tr>
</tbody>
</table>

Table 5-1: Mechanical properties of Si, Al₂O₃, TiO₂ and Parylene-C.
The thickness of the silicon membrane is about \( t_1 = 42 \, \mu m \) given by the manufacturer. It can be easily obtained that the z-coordinate of neutral plane before coating is:

\[
Equation \, 5-10 \\
z_c = \frac{E_i}{1 - v_i^2} t_1 \frac{t_1}{2} = t_1 \frac{t_1}{2}
\]

\[
Equation \, 5-11 \\
z_c = \frac{42 \, \mu m}{2} = 21 \, \mu m
\]

So the neutral plane is in the middle of the silicon membrane, as it was expected from symmetry.

For two layer plate equation 5-5 can be written as the following:

\[
Equation \, 5-12 \\
z_c = \frac{E_i}{1 - v_i^2} t_1 \frac{t_1}{2} + \frac{E_2}{1 - v_2^2} t_2 \left( t_1 + \frac{t_2}{2} \right)
\]

The \( z_c \) for each of the sensors after coating can be easily obtained by means of equation 5-9. The calculation results are listed in Table 5-2.

<table>
<thead>
<tr>
<th>Coating</th>
<th>Not coated</th>
<th>41nm TiO₂</th>
<th>100nm TiO₂</th>
<th>49.8nm Al₂O₃</th>
<th>100nm Al₂O₃</th>
<th>5μm Parylene-C</th>
<th>20μm Parylene-C</th>
</tr>
</thead>
<tbody>
<tr>
<td>( z_c ) (( \mu m ))</td>
<td>20</td>
<td>20.03</td>
<td>20.07</td>
<td>20.06</td>
<td>20.13</td>
<td>20.05</td>
<td>20.29</td>
</tr>
<tr>
<td>% change</td>
<td>0%</td>
<td>-0.15%</td>
<td>-0.37%</td>
<td>-0.32%</td>
<td>-0.63%</td>
<td>-0.27%</td>
<td>-1.44%</td>
</tr>
</tbody>
</table>

Table 5-2: expected sensitivity change due to shift of neutral plane after coating

5.3.2 Stiffness Changes Due to Coating

Equation 5-6 for two layers can be rewritten as Equation 5-13:
Equation 5-13: 

\[ D_c = \frac{E_1}{1 - \nu_1^2} \left( \frac{t_1^2}{12} + t_1 \left( \frac{t_1}{2} - z_c \right)^2 \right) + \frac{E_2}{1 - \nu_2^2} \left( \frac{t_2^2}{12} + t_2 \left( \frac{t_2}{2} - z_c \right)^2 \right) \]

So for calculated \( z_c \)'s, moment of inertia can be obtained. Table 5-3 shows the calculated values.

<table>
<thead>
<tr>
<th>Coating Not coated</th>
<th>41nm TiO₂</th>
<th>100nm TiO₂</th>
<th>49.8nm Al₂O₃</th>
<th>100nm Al₂O₃</th>
<th>5μm Parylene-C</th>
<th>20μm Parylene-C</th>
</tr>
</thead>
<tbody>
<tr>
<td>( D_c (Nm)^{+10^-} )</td>
<td>9.01</td>
<td>9.05</td>
<td>9.11</td>
<td>9.10</td>
<td>9.18</td>
<td>9.09</td>
</tr>
<tr>
<td>% change</td>
<td>0%</td>
<td>-0.45%</td>
<td>-1.10%</td>
<td>-0.95%</td>
<td>-1.90%</td>
<td>-0.92%</td>
</tr>
</tbody>
</table>

Table 5-3: expected sensitivity change due to increased stiffness after coating

Total change in sensitivity can be obtained by considering the effect of changes in neutral plane as well as changes in flexural rigidity. From equation 5-7 the total change in sensitivity can be obtained. Results are shown in table 5-4.

<table>
<thead>
<tr>
<th>Coating Not coated</th>
<th>41nm TiO₂</th>
<th>100nm TiO₂</th>
<th>49.8nm Al₂O₃</th>
<th>100nm Al₂O₃</th>
<th>5μm Parylene-C</th>
<th>20μm Parylene-C</th>
</tr>
</thead>
<tbody>
<tr>
<td>% change</td>
<td>0%</td>
<td>-0.60%</td>
<td>-1.47%</td>
<td>-1.26%</td>
<td>-2.53%</td>
<td>-1.19%</td>
</tr>
</tbody>
</table>

Table 5-4: Percentage of total change in sensitivity after coating

### 5.4 Structure of SP82

The pressure sensor used in this project, SP82 by MEMScAP, is a relative differential pressure sensor. It has a silicon substrate with four doped piezoresistors which form a Wheatstone bridge configuration. It also has two resistors for controlling the temperature. The sensor has two pressure ports on bottom and top of the sensor (Figure 5-3), and measures the difference between those two pressures (see Figures 5-4 and 5-5).
Figure 5-3: (a) SP82 pressure sensor with two ports. The size is compared with one Norwegian krone (b) the sensor with removed cap. Silicon die is gold wire bonded to 12 legs.

Figure 5-4: Wheatstone bridge configuration of piezoresistors in SP82

Figure 5-5: SP82. The silicon die with the wire bonds are shown in the picture. The sensor has 12 legs on the bottom which are connected to 12 corresponding pins. 3 pins are not used.

The goal of this project is to study the influence of different coatings with various types and thicknesses on the sensitivity of the pressure sensor. In addition the effect of temperature on the sensor (the temperature coefficient of offset, TCO) is studied. The
sensitivity of six similar sensors was measured by cycling pressure from -500 mbar to 500 mbar at two constant temperatures, 25 and 50 degrees centigrade. The temperature was then cycled from 20 to 55 degrees centigrade with constant pressure of 0 mbar. These measurements were performed on the sensors without coating. Three different biocompatible coatings were then applied to the six sensors, \( TiO_2 \), \( Al_2O_3 \) and \( Parylene-C \). Two different coating thicknesses were used for each coating (see Table 5-5). After coating, the same measurements were repeated for each coated sensor. The compared results are presented in chapter 6.

<table>
<thead>
<tr>
<th>Sensor Number</th>
<th>Coating Type</th>
<th>Thickness</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>TiO(_2)</td>
<td>41nm</td>
</tr>
<tr>
<td>2</td>
<td>TiO(_2)</td>
<td>100nm</td>
</tr>
<tr>
<td>3</td>
<td>Al(_2)O(_3)</td>
<td>49.8nm</td>
</tr>
<tr>
<td>5</td>
<td>Al(_2)O(_3)</td>
<td>100nm</td>
</tr>
<tr>
<td>4</td>
<td>Parylene-C</td>
<td>5(\mu)m</td>
</tr>
<tr>
<td>6</td>
<td>Parylene-C</td>
<td>20(\mu)m</td>
</tr>
</tbody>
</table>

Table 5-5: Sensors numbers, their coating and thicknesses.
6 Experiments and Results

6.1 Measurement set up

The sensor is placed in a stainless steel pressure chamber (Figure 6-1) under controlled condition.

![Stainless steel pressure chamber diagram](image1)

**Figure 6-1**: Stainless steel pressure chamber. The sensor is placed into the chamber. It is connected to the supply and other devices via wires passing through to the chamber cap. The cap is a Plexiglas plate which makes the chamber completely air tight.

Pressure in the chamber is measured and controlled by a DPI 515 (by Druck). The DPI 515 generates a controlled pressure in the pressure chamber by receiving a high (2.5bar) and a low (<0.1bar) pressure on the inlet tube terminals (Figure 6-2).

![Backside view of DPI515](image2)

**Figure 6-2**: backside view of DPI515

The high pressure comes from dry, pressurized Nitrogen gas. The low pressure comes from a vacuum pump that generates a low pressure of dry air. A controlled pressure...
between the high and low pressures is generated by DPI 515 and connected to the chamber. The chamber is completely air tight by means of a Plexiglas plate which is screwed to the stainless steel chamber. There is a Pt100 probe inside the chamber to measure the inside temperature. It is connected to a tempcontrol P600 series with a monitor and also connected to a computer. The pressure chamber is then placed in an open bath circulator DL30-V15/B (by Haake). The bath circulator has ±0.03K accuracy and 0.01K resolutions according to the manufacturer. When the environment and the measuring system have large temperature difference (about maximum 30°C), the measurements will be disturbed due to condensation problems [9]. According to a rule of thumb suggested by metrologist experts stable measurement conditions can be guaranteed by submerging the pressure chamber in water to a certain depth (about 15 times the diameter of Pt100 probe) [9], to avoid the heat leakage from the chamber to the air.

The sensor is connected to a 3V battery and a digital multimeter records the output voltage of the sensor as well as the supply voltage. The multimeter is a low noise 7 ½ digit one by Keithley (Model 2010). All devices are connected to a computer and the condition is controlled by a program in LabVIEW¹ (by National Instruments). Measured values by DPI 515 pressure controller and DL30-V15/B temperature controller are recorded by the computer as well as data sampled by the Pt100 thermometer and the Keithley multimeter.

¹ See Appendix A
Figure 6-3: Measurement set up: (a) Computer, (b) DPI 525 pressure controller/calibrator, (c) Model 2010 Multimeter (Keithley), (d) water bath and circulator DL30-V15/7 (e) Tempcontrol P650 (f) stainless steel pressure chamber with sensor inside.

6.2 Measurement

All sensors have undergone two different measurements: pressure cycle and temperature cycle; once before coating and then after coating. In Figure 6-4 performed experiments are summarized.

Figure 6-4: Experiments flow chart
6.2.1 Pressure Cycle

The pressure sensor (SP82) is put in the chamber while one port has atmospheric pressure and the second pressure port is inside the chamber. Pressure inside the chamber is changed from -500mbar to +500mbar (relative) in 100mbar steps and thereafter from +500mbar to -500mbar with the same step size. In each step the sensor’s output voltage is recorded. The cycle is repeated three times to ensure the sensor’s repeatable performance. These measurements are done in constant temperature once in 25 degrees and then in 50 degrees centigrade.

The results show a very high linearity and repeatability (Figures 6-5 and 6-6). Before coating all the sensors had sensitivity of $0.1\frac{V}{bar}$ at $T=25^\circ C$ and $0.09\frac{V}{bar}$ at $T=50^\circ C$.

Table 6-1 shows the sensitivity and offset for each sensor before coating.

<table>
<thead>
<tr>
<th>Sensor number</th>
<th>Temperature</th>
<th>$T=25^\circ C$</th>
<th>$T=50^\circ C$</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Sensitivity(V/bar)</td>
<td>Offset(mV)</td>
<td>Sensitivity(V/bar)</td>
</tr>
<tr>
<td>1</td>
<td>-0.0962</td>
<td>-0.278</td>
<td>-0.0908</td>
</tr>
<tr>
<td>2</td>
<td>-0.0992</td>
<td>+1.817</td>
<td>-0.0937</td>
</tr>
<tr>
<td>3</td>
<td>-0.0993</td>
<td>-0.512</td>
<td>-0.0932</td>
</tr>
<tr>
<td>4</td>
<td>-0.0989</td>
<td>-1.231</td>
<td>-0.0932</td>
</tr>
<tr>
<td>5</td>
<td>-0.101</td>
<td>+1.034</td>
<td>-0.0950</td>
</tr>
<tr>
<td>6</td>
<td>-0.102</td>
<td>+1.782</td>
<td>-0.0963</td>
</tr>
</tbody>
</table>

Table 6-1: Measured Sensitivity and offset of each sensor before coating

\[ Sensitivity_{avr.\ 25^\circ C} = -0.0994\frac{V}{bar} \quad Sensitivity_{avr.\ 50^\circ C} = -0.0937\frac{V}{bar} \]
\[ Offset_{avr.\ 25^\circ C} = 0.3293\text{mV} \quad Offset_{avr.\ 50^\circ C} = 0.3588\text{mV} \]
Pressure Measurement (in 25 deg. C)

![Pressure Measurement Graph](image)

\[ y = -1 \times 10^{-4}x + 0.0018 \]

\[ R^2 = 1 \]

**Figure 6-5:** Pressure cycle for one of the sensors before coating. The results show high linearity.

**Repeatability**

![Repeatability Graph](image)

**Figure 6-6:** The output voltage shows a good repeatability in three cycles. In each pressure step output voltage is sampled 4 times. Points \(a_1\) to \(a_3\) correspond pressure of 500mbar and points \(b_1\) to \(b_4\) correspond pressure of \(-500\)mbar.

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After having the sensors coated, same pressure measurements are done. Results show a good linearity and repeatability (see Figures 6-7 and 6-8). Table 6-2 shows these results.

<table>
<thead>
<tr>
<th>Sensor number</th>
<th>Sensor coating</th>
<th>$T=25^\circ C$</th>
<th>$T=50^\circ C$</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Sensitivity(V/bar)</td>
<td>Offset(mV)</td>
<td>Sensitivity(V/bar)</td>
</tr>
<tr>
<td>1</td>
<td>41nm TiO$_2$</td>
<td>-0.0962</td>
<td>-0.121</td>
</tr>
<tr>
<td>2</td>
<td>100nm TiO$_2$</td>
<td>-0.0985</td>
<td>+2.153</td>
</tr>
<tr>
<td>3</td>
<td>49.8nm Al$_2$O$_3$</td>
<td>-0.0976</td>
<td>-0.277</td>
</tr>
<tr>
<td>4</td>
<td>5µm Parylene-C</td>
<td>-0.0982</td>
<td>-1.160</td>
</tr>
<tr>
<td>5</td>
<td>100nm Al$_2$O$_3$</td>
<td>-0.0980</td>
<td>-3.034</td>
</tr>
<tr>
<td>6</td>
<td>20µm Parylene-C</td>
<td>-0.0992</td>
<td>+0.371</td>
</tr>
</tbody>
</table>

Table 6-2: Measured Sensitivity and offset of each sensor after coating

Pressure measurement(in 25 deg. C)

\[
y = -1E-04x + 0.0022 \\
R^2 = 1
\]

Figure 6-7: Pressure cycle for one of the sensors after coating. It is coated with 100nm thick TiO$_2$. The results show high linearity
**Figure 6-8:** The results of three pressure cycles of a sensor after coating shows good repeatability. The coating has not influenced the repeatability of the sensor.

In Figure 6-9 the voltage-pressure of all sensors before and after coating is shown. Linear performance of all sensors can be clearly seen in this figure.

**Figure 6-9:** Pressure cycle of all sensors before and after coating
<table>
<thead>
<tr>
<th>Sensor Number</th>
<th>Change in Sensitivity</th>
<th>Calculation result</th>
</tr>
</thead>
<tbody>
<tr>
<td></td>
<td>Experimental result at 25°C</td>
<td>Experimental result at 50°C</td>
</tr>
<tr>
<td>1</td>
<td>0.05%</td>
<td>-0.05%</td>
</tr>
<tr>
<td>2</td>
<td>-0.73%</td>
<td>0.00%</td>
</tr>
<tr>
<td>3</td>
<td>-1.75%</td>
<td>-0.77%</td>
</tr>
<tr>
<td>4</td>
<td>-0.70%</td>
<td>0.04%</td>
</tr>
<tr>
<td>5</td>
<td>-3.18%</td>
<td>-2.49%</td>
</tr>
<tr>
<td>6</td>
<td>-2.78%</td>
<td>-0.94%</td>
</tr>
</tbody>
</table>

Table 6-3: comparison of variation in sensitivity calculation and experimental results

Figure 6-10: Sensitivity of sensors before and after coating is shown in blue and pink dots respectively. The pink dots are below the blue dots which mean that the coating has decreased the sensitivity. Sensor number 1 and 2 are the sensors with a layer of TiO₂ with thickness of 41 and 100 nm respectively. The change in sensitivity in the sensor with thicker layer at 50°C is greater than other sensors coated with titanium-dioxide. Sensor number 3 and 5 are coated with thin and thick Alumina. Thick alumina has the largest variation at both 25 and 50°C. Sensor number 4 which has a 5µm thick layer of Parylene-C does not have big change in sensitivity. Sensor number 6 with a 20µm thick Parylene-C has a relatively large decrease in sensitivity at 25°C.

The changes in sensitivities are also shown in Figure 6-10. Coated sensors have lower sensitivity. In the sensor coated with TiO₂ the changes are much smaller than what was predicted by the calculations (see table 6-3). In sensors coated with Alumina changes in sensitivity are in approximate agreement with the calculation. In sensor number 6 which
is coated with 20µm Parylene-C, calculations expected bigger change in sensitivity. Although sensitivity had decreased after coating, the calculated decrease in sensitivity is found to be bigger than the measured decrease.

**Figure 6-11**: changing rate of sensitivity in 6 sensors. Sensor number 5 which is coated with 100nm thick Alumina and then sensor number 6 coated with 20µm thick Parylene-C have bigger changes.
The offsets in voltage-pressure equations are compared. In most of the sensors offset does not have large change after being coated except in sensors number 5 and 6. These changes are shown in Figure 6-12.

![Graph showing offset difference](image)

**Figure 6-12:** The changes in offset in sensor number 5 and 6 in both 25°C and 50°C are much larger than other sensors.

### 6.2.2 Temperature Cycle

Temperature coefficient of offset for each sensor is measured. The procedure is to vary the temperature in a constant pressure. The temperature is changed between 20 and 55 degrees centigrade while the relative pressure between the two ports of pressure sensor is zero. The output voltage of the sensor is recorded. The same measurement is done for the sensors after coating (see Figure 6-13).

The offset of thinner layers are less than thicker layers of same type of material. For the sensor coated with 100nm Alumina this offset is much greater than for other sensors. This is discussed below. For the sensors coated with Parylene-C (sensor number 4 and 6) the graphs are less linear and have some hysteresis. Hysteresis and nonlinearity is bigger in the sensor with thicker Parylene-C coating (Figure 6-13).
Temperature Cycle

**Figure 6-13:** Temperature cycle of all sensors before and after coating. Sensor 6 with a 20µm parylene-C film has the most nonlinear behavior (see line marked as S6_C). The average voltage in sensor 5 with 100nm thick Alumina layer has the biggest difference with the average voltage of same the sensor before coating.

Hysteresis has been approximately calculated for all sensors. The voltage-temperature path is approximated to be linear both for increasing and decreasing temperature. The area surrounded between the two lines, $A$ in $\Delta T = T_2 - T_1$ is calculated. This area is approximately equal to hysteresis (Figure 6-14).

**Figure 6-14:** Approximation for hysteresis. The calculated hysteresis is equal to the area $A$ in this picture which is equal to $A = A_2 - A_1$ while $A_1 = \frac{1}{2}(v_1 + v_2)(T_2 - T_1)$ and $A_2 = \frac{1}{2}(v_2 + v_3)(T_2 - T_1)$. 

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Calculated hysteresis for all sensors are listed in Table 6-4. Sensors coated with TiO$_2$ do not have large hysteresis compared to Parylene-C and Alumina. Changes in hysteresis are directly proportional to thickness in these two types of coating. Figure 6-15 shows these changes schematically.

The temperature coefficient of offset for each sensor is calculated. Table 6-4 shows the results. The maximum TCO is less than 8µV/°C. When the sensitivity is assumed 0.1V/bar, 8µV/°C corresponds to offset less than 0.1mbar per degree centigrade. The largest variation in TCO belongs to the sensor coated with 100nm thick alumina. These changes are shown in Figure 6-16. TCO in Parylene-C coated sensors does not depend on the coating’s thickness but in TiO$_2$ TCO has increased when the coating’s thickness is increased.

| Sensor Number | Hysteresis uncoated sensor (mV/°C) | Hysteresis coated sensor (mV/°C) | TCO uncoated sensor (µV/°C) | TCO coated sensor (µV/°C) | |ΔTCO| (µV/°C) | Coating material and thickness |
|---------------|------------------------------------|----------------------------------|----------------------------|--------------------------|----------------|-----------------------------|
| 1             | 0.24                               | -0.41                            | 0.78                       | 0.14                     | 0.6           | 41nm TiO$_2$                |
| 2             | 1.62                               | 1.06                             | 0.51                       | -0.58                    | 1.1           | 100nm TiO$_2$               |
| 3             | -1.74                              | 0.13                             | 5.05                       | 4.06                     | 1.0           | 49.8nm Al$_2$O$_3$          |
| 4             | 0.10                               | 3.90                             | 4.05                       | 2.07                     | 2.0           | 5µm Parylene-C              |
| 5             | -0.37                              | -4.97                            | 0.53                       | 6.55                     | 6.0           | 100nm Al$_2$O$_3$           |
| 6             | -1.58                              | 11.69                            | -1.33                      | 0.61                     | 1.9           | 20µm Parylene-C             |

Table 6-4: Hysteresis and temperature coefficient of offset for all sensors. The sensors coated with Parylene-C have the largest hysteresis. This is increasing by increasing coating thickness. TCO in all sensors is less than 7µV/°C. By taking the average value of 0.1 V/bar for sensitivity this corresponds to less than 1µbar/°C.
Figure 6-15: rate of changes in hysteresis. Sensors coated with Parylen-C have a significant positive hysteresis while thick Alumina coated sensors have negative hysteresis. The hysteresis in sensors coated with TiO$_2$ is not significant.

Figure 6-16: Changes in TCO after coating. Titanium-oxide has the smallest effect, then parylene-C and the thicker Alumina has the largest change coating.

### 6.2.3 Measurement Challenges in Thick Alumina Coated Sensor

Sensor number 5, coated with 100nm thick Alumina, had three broken connections after coating. This was repaired by means of soldering to 3 other pins as shown in Figure 6-17. The large offset can be due to the different temperature properties of various materials.
used for the reparation such as tin used for soldering, plastic insulation of the new wire, etc.

**Figure 6-17:** Sensor number 5 which is coated with 100nm thick Alumina. The disconnections after coating is repaired by means of blue wire soldered to 2 other pins which were not used. Pin 10 and pin 7 are connected to pin 3 and 4 respectively. There is also a tin-bridge between pin 5 and 6 to repair the detached leg on pin number 5.

### 6.3 Temperature Resistor

There is an integrated temperature resistor in the sensor. For one sensor the temperature cycle has been repeated while the temperature resistor was connected to an ohm-meter. A time delay in measuring temperature by Pt100 with respect to actual temperature of the sensor measured by temperature resistor can cause a hysteresis in the TCO curve (even for elements without hysteresis). In Figure 6-18 time-lag between the temperature measured by the resistor integrated in the sensor and that measured by Pt100 is shown. In this figure there is no significant time-lag. It is logical to conclude that the coatings caused the hystereses in the coated sensors. Figure 6-19 shows the relation between temperatures measured by resistor on the sensor and that measured by Pt100. There are traces of hysteresis in this graph. Although it is more accurate to use the temperature resistor on the sensor for measuring sensor's temperature, the temperatures measured by Pt100 are close enough to the actual temperature. This increases the confidence in the Pt100, and indicates that it has been accurate in measuring the sensor ambient temperature.
Figure 6-18: temperature measurements of the temperature resistor and the Pt100 versus time.

Figure 6-19: The linear relation between measured temperature by Pt100 and measured resistance by the ohm-meter. There is a small hysteresis in this graph.
7 Conclusion and Future works

7.1 Conclusion

The effect of different biocompatible coatings on the sensitivity of pressure sensors has been evaluated. Results show that the sensitivity is inversely proportional to the thickness of the coating. For the sensors coated with Titanium-dioxide, this dependence is not significant even when the thickness is 100nm. The most considerable change in sensitivity between six different coatings belongs to 100nm thick alumina which shows a reduction of about 3%. This coating also has a larger effect on sensor offset. Temperature coefficients of offset for the six pressure sensors before and after coating are measured. Parylene-C has the most non-linear behavior with hysteresis in temperature cycles compared with Al₂O₃ and TiO₂. This non-linearity is increased when the coating is thicker.

It seems reasonable to conclude that titanium-oxide has the best properties as coating for pressure sensor with minimum change in sensitivity and thermal properties.

7.2 Future Works

Before such systems can be implanted they must be tested in vitro and in vivo. First the chemical stability of the coatings has to be tested. One idea is to put the sensors into an environment similar to body fluid, and evaluate their performance. It would be interesting to see if the sensor still works after such treatment and also to see if the coatings can withstand the aggressive environment.

If the coated sensor has to be used with ionic solution (body fluid) around it, a way to keep the coated sensor dry is probably necessary. Otherwise the coating will act as a dielectric in a capacitor, and suffer a break-down when exposed to a few volts per 50nm. This pressure sensor was very large to be an implant. The system should be miniaturized and appropriately packaged to become ready for implantation. A communication unit is needed to do the data transmission from sensor to doctor’s computer. The system demands a miniature supply unit suitable for implantation.
8 References


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9 Appendix A

The block diagram of the LabVIEW program used for the measurement in this project is shown.

<table>
<thead>
<tr>
<th>1-A</th>
<th>2-A</th>
<th>3-A</th>
</tr>
</thead>
<tbody>
<tr>
<td>1-B</td>
<td>2-B</td>
<td>3-B</td>
</tr>
</tbody>
</table>

Instruction: Fig.2 to Fig.7 show the block diagram of the LabVIEW program. The instruction above shows how they are connected.
Fig. 3: Block diagram 2-A
Fig. 4: Block diagram 3-A