PRESSURE, FLOW AND OXYGEN SATURATION SENSORS ON ONE CHIP FOR USE IN CATHETERS.

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ABSTRACT

This paper presents a combination sensor chip, which is to be fitted to a catheter for use in intervention therapy. The chip contains an absolute pressure sensor, a thermal flow sensor and a colour sensor to determine respectively: blood pressure, blood flow velocity and oxygen saturation in one location at the same time.

The sensors where fabricated using epi-micromachining. Change in absorption in wavelengths 660nm and 800nm due to the oxygenation of the blood are measured using two stacked photodiodes that serve as a colour sensor.

INTRODUCTION

In the medical fields of intervention radiology, cardiology and neurology, the blood vessels are used as a pathway for catheters to reach places inside the body that need treatment. This method of treatment minimises damage to healthy tissue and reduces discomfort of the patient. However, this indirect access limits the information available to the doctor to what can be seen using fluoroscopy and the pressure transmitted through the hollow catheter. To provide the doctor with more information, sensors are needed to gather data for diagnosis, monitoring of the procedure and checking results. To obtain information on the general state of the blood supply, a combination of sensors was developed to measure the blood pressure, blood flow velocity and oxygen saturation.

This sensor needs to be small enough to be fitted in a catheter with an outer diameter of 1.67 mm (5 French). Additionally, the section of the catheter where the sensor chip is placed will be rigid, where the rest of the catheter is flexible. If the length of the stiff section is too long, the catheter will not be able to follow the twists and turns of the blood vessels. As there also needs to be room for the guide wire, the overall dimensions are limited to 1 mm by 10 mm. Figure 1 shows a schematic of the catheter with the blood sensor.

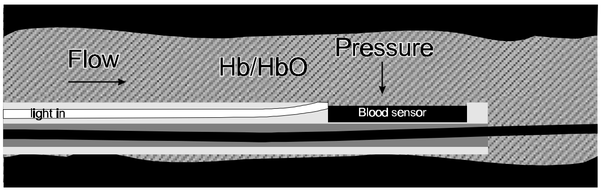

Figure 1. Schematic of the catheter showing the sensor chip in the outside wall leaving enough space for the guide wire.

For a more thorough analysis of the requirements of the sensors and discussion of both the pressure and flow sensors see [1].

PRESSURE SENSOR

The pressure sensor consists of a membrane, formed in 4 µm epitaxially grown polysilicon on a silicon oxide sacrificial layer. This oxide is removed through small etch holes in the membrane, leaving a small cavity underneath (see Figure 2). The holes are then sealed using a silicon oxide, giving a reference pressure of 250 mTorr (3.35 kPa). The deflection of the membrane is measured using polysilicon piezo-resistors forming a Wheatstone bridge. This pressure sensor is robust and planar and will be insensitive to material from the blood deposited on top of it.

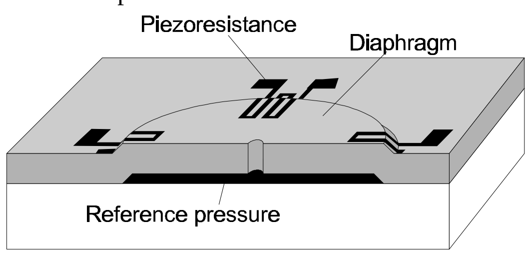

Figure 2. Cross sectional view of the absolute pressure sensors.

FLOW VELOCITY SENSOR

The flow velocity is measured using the thermotransfer principle. The blood is locally heated to two degrees above ambient, and the temperature rise is measured upstream and downstream of the heater. The heater

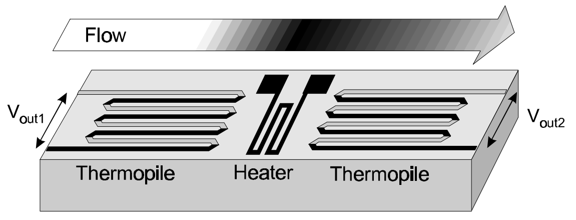

Figure 3. Principle of the thermal flow velocity sensor.

consists of a polysilicon resistor and the temperature difference is measured using polysilicon - aluminium thermopiles (see Figure 3). To reduce the loss thermal loss to the substrate, the flow sensor is placed on the same epi-poly membrane used for the pressure sensor, with the sacrificial oxide still in place.

Materials deposited on the heater or thermopile during use will slow the device down but will not inhibit the working of the sensor.

OXYGEN SATURATION MEASUREMENT

The protein haemoglobin in the blood transports oxygen from the lungs to all parts of the body. It can bind to oxygen and changes to oxyhaemoglobin. The ratio between haemoglobin and oxyhaemoglobin is called the oxygen saturation of the blood and can be used to indicate whether the blood supply is still able to supply sufficient oxygen to the surrounding tissue.

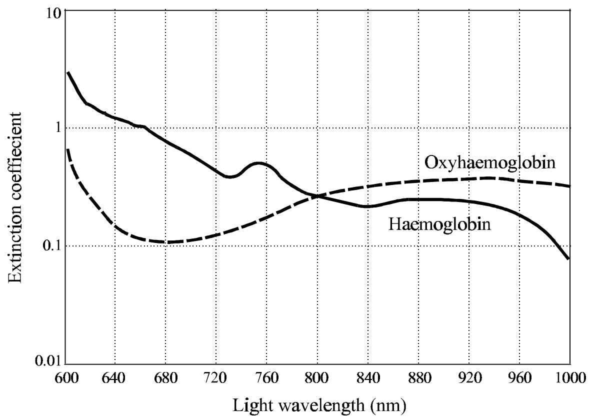

Figure 4. Extinction coefficient as a function of the wavelength for both haemoglobin and fully saturated oxyhaemoglobin

The oxygen saturation sensor uses an established method known as oximetry. Depending on the number of oxygen molecules bound, the absorption of light changes, i.e. it becomes more red when more oxygen is present. As shown in Figure 4, the change is most obvious in light of wavelengths of 660 nm and above 840 nm (940 nm is often used). At 800 nm the absorption is the same for both oxygenated and deoxygenated blood. By using the 800 nm wavelength to establish a reference level of absorption the sensor can be made insensitive to variations in the incoming light, caused by changes in the light sources or in the absorption of material deposited on the surface of the sensor. The response at 660 nm can then be used to determine the level of oxygenation of the blood. Both these measurements can be done using silicon sensors.

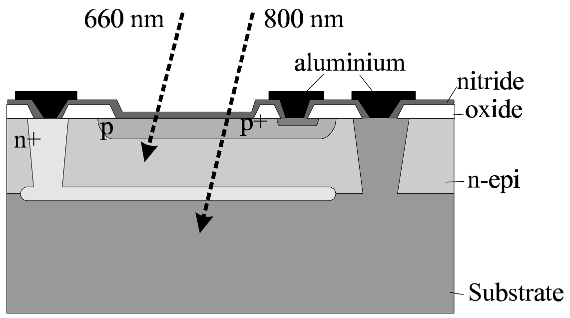

Figure 5. Cross sectional view of the oxygen saturation sensor.

The sensor consists of two vertically stacked photodiodes to keep the sensor area small (see Figure 5) to form a colour sensor [2]. Due to the wavelength dependence of the absorption depth of photons in silicon the infrared light penetrates further and the signal of the deeper photodiode will give a signal due to the infrared light and the top photodiode a signal due to both the red and infrared light. From these two signals the absorption in both wavelengths can be determined and the oxygen saturation calculated. The light used to do the measurements will be applied through an optical fibre in the catheter, illuminating the blood, as there is no room in the catheter for light sources such as LEDs.

PROCESSING

The process used for the fabrication of the sensors is an adaptation of the process developed by P.T.J. Gennissen et. al. [3]. It modifies a standard bipolar electronic process (DIMES-01 [4]) by depositing a sacrificial oxide and a polysilicon seed layers before the growth of the epitaxial layer. This provides us with a 4 µm thick epitaxially grown polysilicon layer on a sacrificial oxide, as the membrane for the pressure and flow sensors and a planar surface. After covering the epi-poly with silicon nitride, the rest of the processing can continue as normal, forming the photodiodes. This also makes it possible to integrate electronics for signal conditioning, on-chip testing, calibration, multiplexing and/or A-to-D conversion to reduce the number of wires needed in the catheter. Before the aluminium interconnect is deposited holes are etched using RIE, and the sacrificial layer underneath the pressure membrane is removed using BHF. The holes are then filled with sequential depositions of silicon oxide. A polysilicon layer is deposited to form the piezo resistors and the thermopiles and the aluminium interconnect is deposited. Finally, the whole chip is covered by silicon nitride to protect it from the blood.

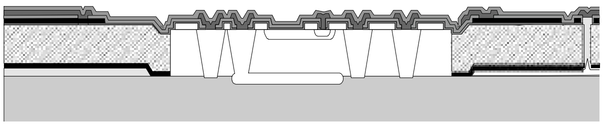

Figure 6. Cross sectional view of the sensor, showing the flow sensor on the right (one end of a thermo-couple), the oxygen saturation sensor in the middle and the pressure sensor on the right.

A cross-section of the whole chip is shown in Figure 6. Part of the pressure sensor is shown on the right and part of the thermal flow sensor on the left. The Oxygen saturation sensor is in the middle.

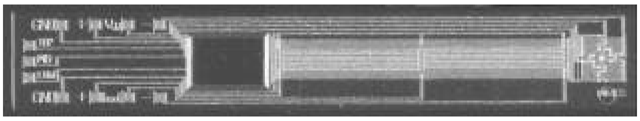

Figure 7. Photograph of the blood sensor with the pressure sensor on the right, the flow sensor in the middle and the oxygen saturation sensor on the left.

The SEM in Figure 7 shows the resulting blood sensor chip. The chip is 1 mm wide and 7 mm long and contains the pressure sensor, the flow sensor and the oxygen saturation sensor.

MEASUREMENTS

The sensors were functionally tested and characterised. The response of the pressure sensor is shown in Figure 8. The flattening of the response above 300 kPa is caused by the compression of the gasses in the cavity, due to it's small size. Above 350 kPa the membrane touches the substrate which serves as an overpressure protection.

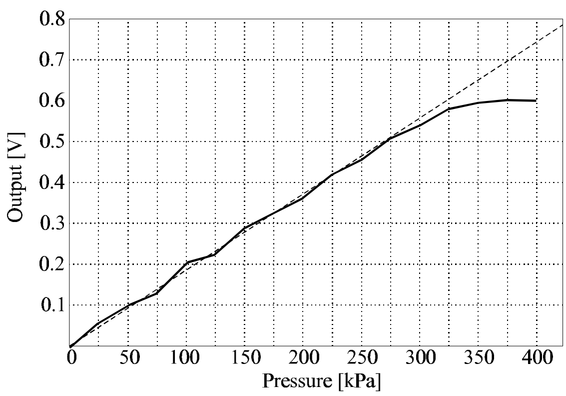

Figure 8. Respons of the pressure sensor.

The pressure sensor showed considerable drift, which settled after several days. This is thought to be the result of stress and the constant loading of the polysilicon membrane and polysilicon piezo-resistors. The response of the flow sensor is shown in Figure 9.

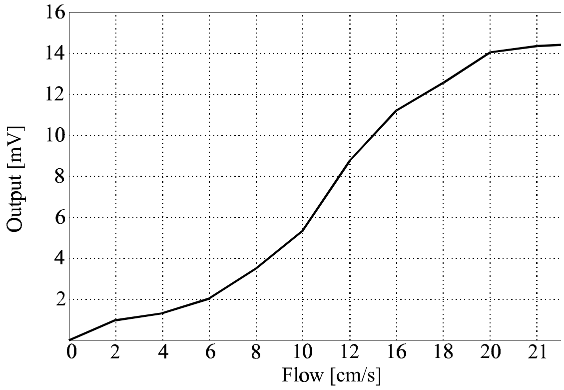

Figure 9. Response of the flow velocity sensor in airflow.

The photodiodes of the oxygen saturation sensor were characterised by determining the wavelength dependence of the response and the dark current. A monochromator was used as a light source. Figure 10 shows the response of the upper and lower diodes as a function of the wavelength of the incoming light. Figure 11 shows the dark current of the upper and lower diodes as a function of the reverse-bias voltage.

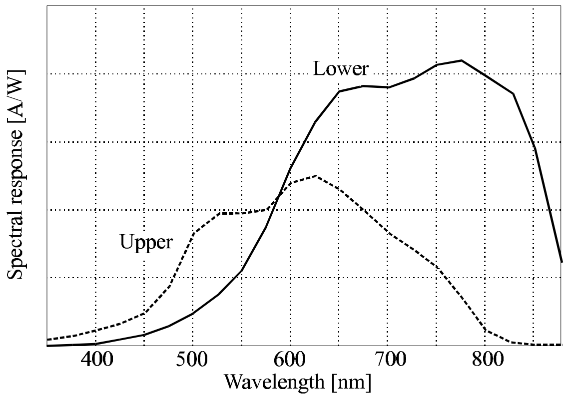

Figure 10. The relative output current of the upper and lower photodiodes as a function of the wavelength of the incoming light.

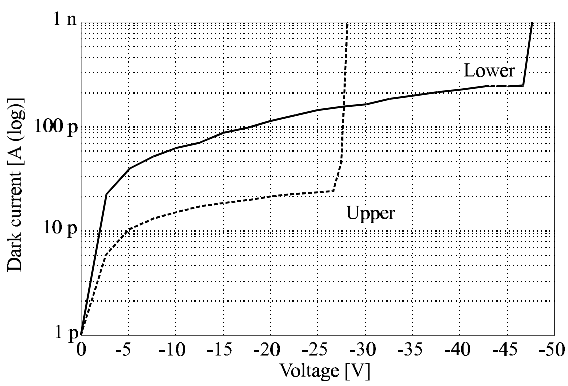

Figure 11. Dark current of the upper and lower photodiodes as a function of the reverse-bias voltage.

As can be seen from these curves the use of standard implantations results in peak responses at slightly lower wavelengths then would be optimal. However, the spectral response is sufficient to discriminate between the two wavelengths and determine the oxygen saturation.

CONCLUSIONS

Using epi-micromachining combined with a standard bipolar electronic process, a combination of a pressure, flow velocity and oxygen saturation sensor was made. The chip is small enough to fit in a 1.67mm diameter catheter, the materials are bio-compatible and the sensors insensitive to the hostile environment that is the human body.

Although compromises had to be made concerning the performance of the sensors, all three sensors function as expected and the photodiodes have a frequency response that allows the determination of the oxygen saturation. However, testing of the sensor in blood has not yet been possible. Initial drift of the pressure signal caused by stress, and loading of the membrane, settled after several days.

ACKNOWLEDGEMENTS

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