A miniaturized pressure sensor with inherent biofouling protection designed for in vivo applications

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***Abstract*—The design, fabrication, and measurement results for a diaphragm-based single crystal silicon sensor element of size 820 µm × 820 µm × 500 µm are presented. The sensor element is designed for *in vivo* applications with respect to size and measurement range. Moreover, it is optimized for long-time operation in the human body through a built-in protection preventing biofouling on the piezoresistors. The sensitivity is about 20 mV/V for a change from 500 to 1500 mbar absolute pressure. This result is comparable to conventional sized micromachined pressure sensors. The output signal is not found to be influenced by exposure to 60 °C for three hours, a normal temperature load for a typical sterilization process for medical devices (Ethylene Oxide Sterilization). The hysteresis is low; < 0.25% of full scale output signal. The sensor element withstands an overload pressure of 3000 mbar absolute pressure. Observed decrease in the output signal with temperatures and observed nonlinearity can easily be handled by traditional electronic compensation techniques.**

I. Introduction

Small-sized, lightweight, and low power-consuming MEMS present new possibilities for monitoring of physiological parameters *inside* the human body. Monitoring of pressure is in general highly important in clinical practice and medical research. Blood pressure, eye pressure, pressure inside the bladder and pressure within big joints are only some of the pressures being routinely measured. Reliable *in vivo* sensors are now being used during surgery or for a short period after surgery, or for immediate inspection [1], [2]. Only few devices, however, are meant for permanent implantation. Reliable *in vivo* pressure sensors are therefore strongly desired by the medical community. For permanent or long-term implantation in the human body, biocompatibility is one major obstacle for the success [3]. Biocompatibility comprises the impact of the implanted device on the body (toxicity, inflammation, wear and degradation etc.), as well as the body's reaction to the implanted device (rejection, encapsulation, biofouling etc.) [4]. For sensors used *in vivo,* corrosion caused by aggressive body fluids and accumulation of biological matter, such as proteins, cells and other biological material (i.e. biofouling), may alter the sensor characteristics and thereby sensor stability. Some of the immunologic responses can be overcome by a smart sensor design. In this paper we present a novel miniaturized, diaphragm-based, piezoresistive, pressure sensor element where the piezoresistors will be unaffected by accumulation of biological matter. With a piezoresistive principle of measurement, the necessary electronics might be separated from the sensor element and placed at a location where the space restriction is less severe. This solution also provides for easier transfer of energy to, and data from, the sensor element through near-field or far-field transmission [5]. The presented design will therefore fit well into a catheter inserted into a body cavity. In this paper the sensor element design, fabrication process, and results from electrical measurements and pressure characterizations are being presented.

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II. Materials and methods

A. Sensor Element Design

Size restriction is the overriding design issue for an implantable medical pressure sensor. Pressure range, sensitivity and stability are other important qualities. To ensure long-term stability, protection from humid, biological surroundings is crucial. The device presented here is a piezoresistive, single crystal silicon sensor element, where pressure is determined by deflection of a diaphragm. Traditionally, surface micromachining technology is the preferred method for making small and planar structures [6]. However, bulk micromachining technology takes the advantage of higher piezocoefficients in single crystalline silicon compared to surface micromachined polysilicon. With innovative use of bulk micromachining technology, we have managed to manufacture a sensor element with high sensitivity, inherent protection of the piezoresistors, and with outer lateral dimensions of only 820 µm x 820 µm.

Altogether 12 550 sensor elements, distributed on five slightly different designs, are present on one wafer. The design is unique with respect to protection of the piezoresistors without diminishing the sensitivity; the piezoresistors are placed at the diaphragm surface facing the vacuum reference cavity closed by the anodic bonded glass wafer. The piezoresistors are therefore not in contact with the biological environment (Fig. 1). Furthermore, a protective coating can be added to the diaphragm without moving the piezoresistors closer to the neutral plane of the diaphragm [7]. A decrease in sensitivity due to the coating is therefore reduced.

The chip outline is shown in Fig. 2. Four p-type piezoresistors form a full Wheatstone bridge configuration. The dashed line shows the resulting etched glass cavity defining the pressure sensitive diaphragm.

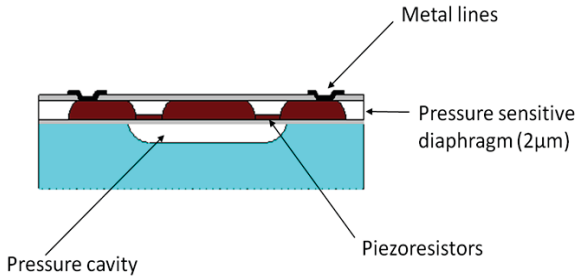

Fig. 1. A schematic cross section of the pressure sensor element with piezoresistors placed at the lower surface of the diaphragm, facing the closed vacuum reference cavity.

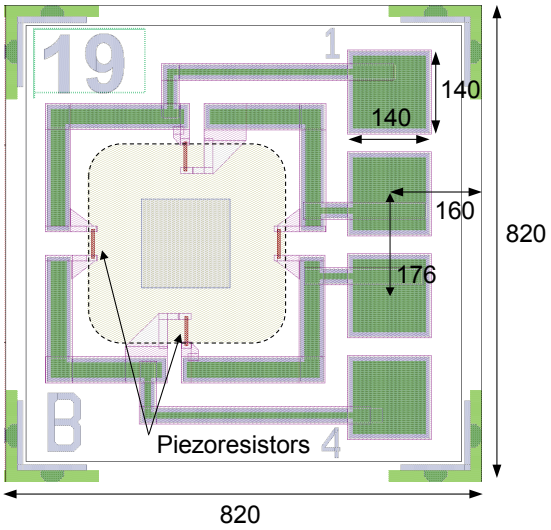

Fig. 2. Chip outline (all measures in µm). The dashed line marks the etched glass cavity defining the pressure sensitive diaphragm. The numbers and letters are for chip identification purposes.

B. *Fabrication process*

The sensor elements were manufactured on 150 mm silicon-on-insulator (SOI) wafers. The thickness of the handle wafer was 625 µm, the buried oxide layer 2000 Å and the device layer was 2 µm. The conductors (boron) were implanted through a 1.2 µm Al mask. The implant energy was 150 keV and the implant dose was 5×10^{15} cm^{-2}. The p-type piezoresistors (boron) were implanted through a resist mask. The implant energy was 15 keV and the implant dose was 1.2×10^{14} cm^{-2}. Bond pads and metal lines were made by Al sputtering. The SOI wafer was anodic bonded to a glass wafer of thickness 525 µm. Pictures of the final miniaturized single pressure sensor element are shown in Fig. 3. The process flow is illustrated in Fig. 4.

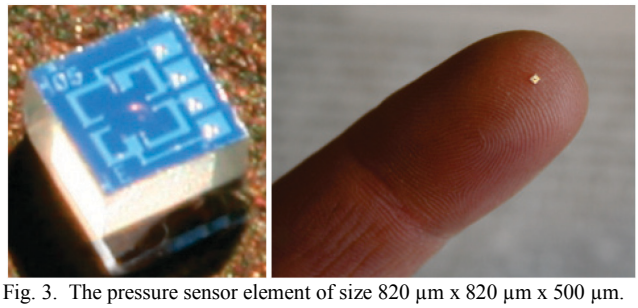

Fig. 3. The pressure sensor element of size 820 µm x 820 µm x 500 µm.

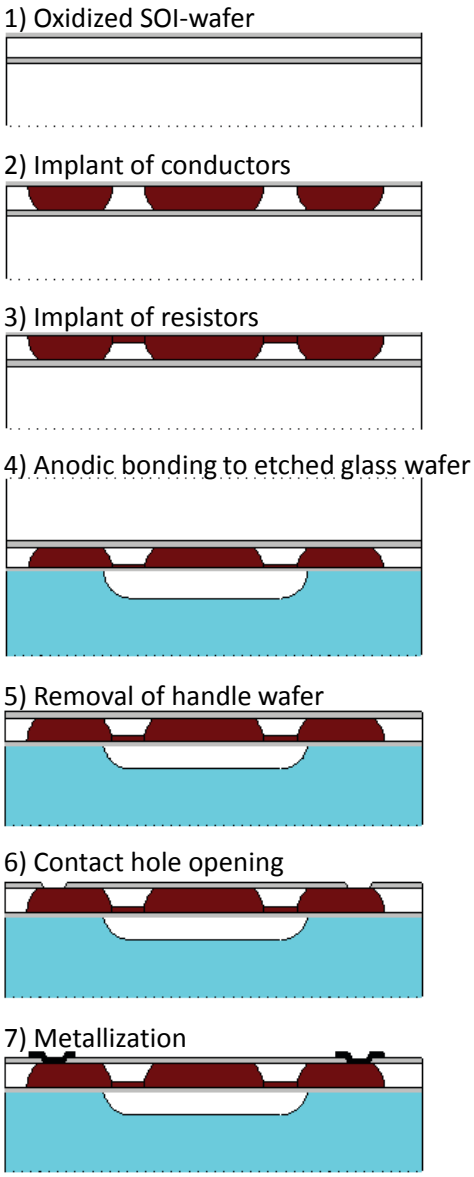

Fig. 4. Illustration of the process flow. Step 1-3 illustrate the processing executed on the separate SOI wafer. The four last steps (4-7) illustrate the processing of the SOI / glass stack. Processing of the separate glass wafer is not shown.

C. *Measurement setup and characterization*

Before dicing, the wafers were electrically probed using a TSK A-PM-90A automatic probe station with a dedicated probe card. The Wheatstone bridge resistances and bridge signal at 5 V and atmospheric pressure were measured. All 12550 dies were measured on wafer number 3, whilst for wafers 1, 2 and 4 approximately 1200 dies across the wafer were measured.

Pressure characterization was performed on sensor elements packaged in ceramic chip carriers, placed inside a pressure chamber. The pressure in the chamber was controlled by a combined pressure controller and calibrator (Druck DPI 515), with nitrogen gas as pressure medium. Pressures below atmospheric pressure were provided by a

sorption pump. The pressure chamber was submerged in a temperature controlled water bath (Haake DL30-V15/B) and a PT100 temperature probe placed inside the chamber recorded the temperature in the proximity of the sensors. The bridge output signal was measured at an excitation voltage of 2.9 V for the pressure range 500 – 1500 mbar absolute pressure (mbara). Measurements were carried out for increasing and decreasing pressures to account for hysteresis effects. Each pressure sweep was done for temperatures from 35 °C to 42 °C. To investigate the influence from a typical sterilization process for medical devices (Ethylene Oxide Sterilization), characterization was repeated after exposure of the sensor element to 60 °C for three hours. Finally the sensor elements were overload tested for pressures up to 3000 mbara.

III. Results

Results from electrical measurements on four different wafers are shown in Table 1. Results from pressure characterization are shown in Fig. 5.

Table 1. Results from electric probing of bridge resistance and output signal at 5 V and atmospheric pressure for four different wafers. Note that the number of measured dies on wafer 3 is larger than for the other three wafers.

Wafer no	*Bridge resistance*		*Output signal*	
	avg. [kΩ]	STD [kΩ]	avg. [mV]	STD [mV]
1	12.4	0.7	154	28
2	11.5	1.5	197	47
3	11.4	0.8	167	60
4	10.2	0.5	261	24
1-4	11.4	1.0	176	61

From Fig. 5 we find that a) The sensitivity is about 20 mV/V for a pressure change of 1000 mbar and the sensor element withstands an overload pressure of 3000 mbara, b) The hysteresis is below 0.25% of full scale output signal (FSO), c) The output signal decreases with increasing temperature, and d) the bridge output signal is not influenced by exposure to 60 °C for three hours.

IV. Discussion

The measured sensitivity for the small-sized pressure sensor element presented here (Fig. 5 a), is comparable to a conventional-sized micromachined pressure sensor [8]. The observed decrease in output signal by increasing temperature (Fig 5 c), can be explained by several effects: i) the rest pressure in the cavity ii) a higher coefficient of expansion for glass than for Si resulting in increasing tension with increasing temperature, and iii) temperature coefficient of sensitivity (TCS) for the piezoresistors, which may all contribute to a lower output signal. However, both the observed temperature dependence and the observed non-linearity can be handled by traditional electronic compensation techniques [9], [10].

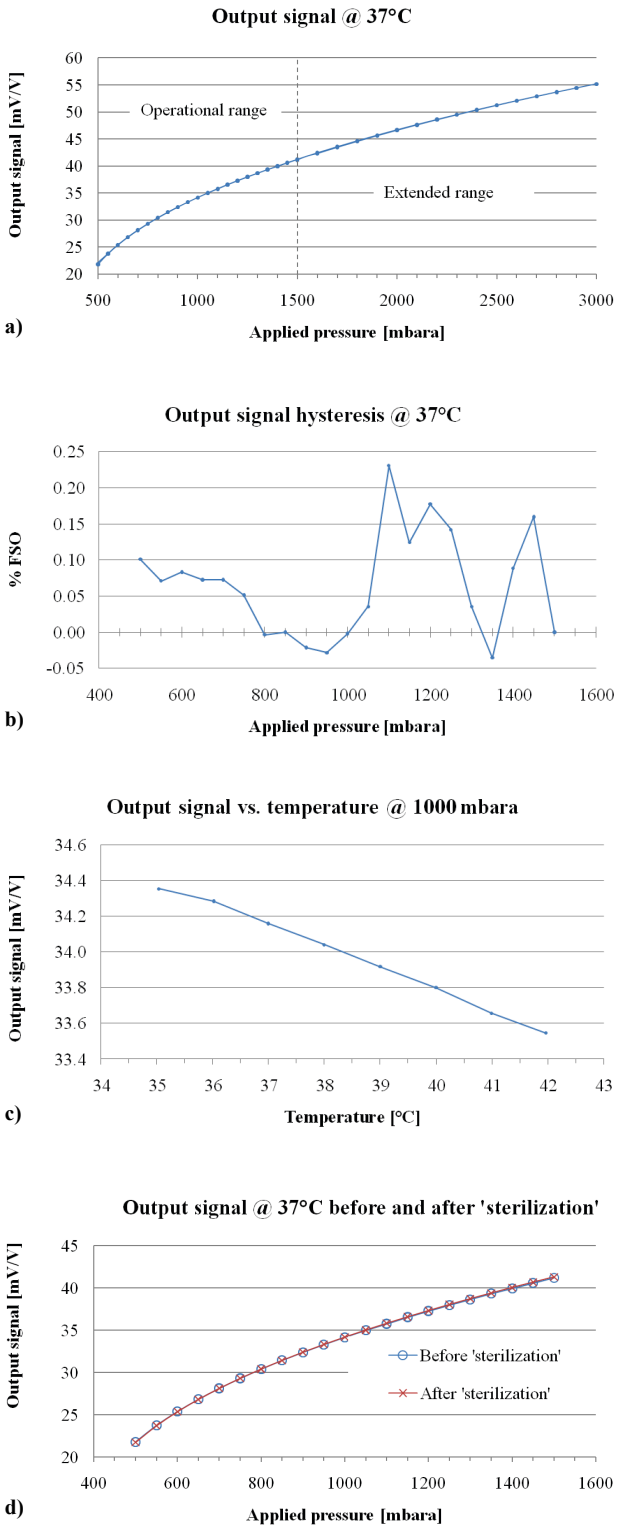

Fig. 5. Results from pressure characterization, typical measurements from one die are shown: a) output signal as a function of pressure, b) signal hysteresis at body temperature, c) temperature shift in the output signal at 1000 mbara, as a function of temperature around body temperature, and d) shift in output signal after the equivalent to a sterilization procedure.

V. Conclusion

The presented miniaturized pressure sensor element shows promising results as a permanent implantable *in vivo* sensor. For a finished packed sensor, however, a solution for protection of the bond pads and bonding wires must be developed. The presented senor element will be used for future *in vitro* and *in vivo* tests to study protective and antifouling coatings. At present, *in vitro* tests in human synovial fluid are commenced. Furthermore in vivo tests in pigs are planned to take place the coming year. Such work is important to ensure long-term operation of pressure sensors in the human body.

Acknowledgments

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