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Design and Optimization of Dual Optical Fiber MEMS Pressure Sensor For Biomedical Applications

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Abstract: A novel Single Deeply Corrugated Diaphragm (SDCD) based dual optical fiber Fabry-Perot pressure sensor for blood pressure measurement is proposed. Both mechanical and optical simulations are performed to demonstrate the feasibility and superior performance of the proposed sensor. Result shows that less than 2% nonlinearity can be achieved for the proposed sensor using optimal Fabry-Perot microcavity. Also, the fabrication process of the proposed sensor is given, instead of complicated fusion bonding process, only bulk and surface micromachining techniques are required which facilitate the mass production of such biocompatible and disposable pressure sensors.

1. Introduction
Pressure measurement of the blood in the coronary artery is most commonly performed using a fluid filled pressure transferring catheter and a pressure sensor external to the body or using a direct measurement at the location inside the coronary artery using a miniaturized catheter based pressure sensor. By measuring the pressure with the miniature sensor inside the artery the time response as well as the accuracy can be improved. The miniature sensor based measurement is well suited for use in balloon angioplasty [1]. Pressure sensors intended for use in medical catheter based intravascular applications must be ultra miniaturized in sizes. Among the proposed techniques, the ability to guide signals to and from a measurement site through an optical fiber, electrical passivity, and immunity to electromagnetic interference make fiber-optic sensor technology attractive for use in biomedical pressure measurements. Furthermore, optical fiber is inexpensive and environmentally friendly, which is important for use in disposable medical equipment.

A fiber-optic pressure sensor [2-5] typically utilizes a sensor head that carries a thin silicon diaphragm in front of the optical fiber end’s surface. However, complicated fusion bonding process is usually required to form the microcavity. This limits the flexibility and the accuracy of device structural dimensions, restricts the compatibility with conventional very large scale integration (VLSI) fabrication technology and leads to low manufacturing yield. In our previous work, we successfully proposed and fabricated a single deeply corrugated diaphragm (SDCD) based optical fiber pressure sensor based on only bulk and surface micromachining techniques [6]. Instead of fusion bonding process, sacrificial layers are etched to form high quality cavities with air gaps. Higher yield and better performance through precisely controlled thickness of the cavity gap can therefore be achieved. On the
other hand, the key novelty marking the proposed SDCD is that it consists of a flat bottom-region that behaves as a normal flat diaphragm, and suspending sidewalls that serve as stress concentrator and buffer, thereby enhancing flatness of the diaphragm under pressurized deflection while reducing temperature dependence by releasing the thermally induced stress. As a result, SDCD based optical fiber pressure sensor can successfully reduce signal averaging effect and thermal cross-sensitivity [7]. However, since intensity-modulated devices measure absolute intensity, intensity losses resulting from source intensity fluctuations or fiber microbend losses may induce phantom pressure during measurement. To minimize such unwanted perturbation in output signal, we propose a novel pressure sensor that relies on the relative intensities of the twin receiving fibers rather than on the absolute intensity of the single one.

2. Proposed dual optical fiber Fabry-Perot pressure sensor

The proposed dual optical fiber based Fabry-Perot pressure sensor are shown in Fig.1, where SDCD act as sensing diaphragm and moving mirror, and the fixed mirror is deposited on the silicon substrate. Air gap is formed by etching the sacrificial layer between moving mirror and fixed mirror and packaging sleeve is fabricated to hold the single-mode optical fibers (9/125 μm). The differential pressure deflects the sensing SDCD through pressure access hole and changes the reflected optical intensities in both optical fibers. Infrared wavelengths of $\lambda_1 = 1300\text{nm}$ and $\lambda_2 = 1550\text{nm}$ are used and we define measurand $I_r$ as:

$$I_r = \frac{I_1 L_1 R_1(\lambda_1)}{I_1 L_1 R_1(\lambda_1) + I_2 L_2 R_2(\lambda_2)}$$  \hspace{1cm} (1)

where $I, L, R(\lambda)$ are the incident optical intensity, optical system losses and absolute reflectance of Fabry-Perot microcavity under different wavelengths for each optical fiber. If incident intensities are adjusted to same and the optical system losses is wavelength independent, such as microbending loss, the $I_r$ become the ratio of absolute reflectances as,

$$I_r = \frac{R_1(\lambda_1)}{R_1(\lambda_1) + R_2(\lambda_2)}$$  \hspace{1cm} (2)

Therefore, the perturbation of measurand induced by intensity loss is successfully removed when dual optical fiber Fabry-Perot pressure sensor is considered. On the other hand, the response curve of $I_r$ is still a periodic function with the applied pressure. Since $I_r$ is the ratio between sinusoidal absolute reflectance, higher linearity can be achieved when compared with the measurement based on absolute reflectance detection.
3. Mechanical design of sensing diaphragm

Thin-film silicon-compatible materials, such as polysilicon and LPCVD (low-pressure chemical vapor deposition) low-stress silicon-rich silicon nitride, have good optical characteristics. They are highly transparent especially under infrared wavelength and the optical absorption of these dielectric films can be omitted. On the other hand, they are biocompatible [8]. Therefore, we proposed a multilayered thin film consisting of two polysilicon layers with one silicon nitride layer (polySi/SiNx/polySi) as sensing diaphragm. Instead of single film layer, multilayered film stack is used to achieve less resultant initial stress through the cancellations of the compressive and the tensile stresses of different individual layers. Since the sensing range of blood pressure is from 0 to 150 mmHg, the thickness of multilayer film stack is set to 1.5 μm based on Ansys simulation. According to schematic view shown in Fig.1, the bottom region of SDCD is set to 150×150μm² and the corrugation depth of SDCD is set to 40 μm. Due to the symmetry of the SDCD, only one quarter of the whole structure is analyzed to reduce model complexity and computation time, as shown in Fig.2. Also, appropriate boundary conditions for nodes on the clamped edge and symmetric boundary conditions for the nodes lying along the boundary between modeled and non-modeled regions are both imposed properly before simulations. The load deflection behavior of proposed sensing diaphragm is shown in Fig.3. It should be pointed out there is static deflection of SDCD prior to any external loading, this is called warpage which is induced by initial stress. In FEM simulations, initial stress of the multilayered diaphragm is assumed to be 70MPa and was induced in the diaphragm by the use of the “cooling temperature” method as proposed by Zhang and Wise [9].

4. Optical structure of Fabry-Perot microcavity for sensing application

The proposed Fabry-Perot microcavity is shown in Fig.4, where the moving mirror is formed by one silicon nitride layer sandwiched between two polysilicon layers (polySi/SiNx/polySi) and the stationary mirror is silicon nitride layer (SiNx).
Based on Macleod characteristic matrix technique [10], the absolute reflectance of Fabry-Perot microcavity can be calculated as,

\[
R = \left( \frac{\eta_0 B - C}{\eta_0 B + C} \right)^* \left( \frac{\eta_0 B - C}{\eta_0 B + C} \right), \quad \text{where } \left[ \frac{B}{C} \right] = \left( \prod_{i=1}^{s} \left[ \cos \delta_i \frac{i \sin \delta_i}{\eta_i \cos \delta_i} \right] \right) \left[ \frac{1}{\eta_s} \right]
\]

\[
\delta_i = \frac{2 \pi n_i d_i \cos \theta_i}{\lambda}, \quad n_i, \eta_i, d_i, \theta_i \text{ are the refractive index, the optical admittance, the physical thickness and the angle of incidence of the } \text{ith thin film layer in the multilayered mirror respectively, while } \eta_s \text{ and } \eta_0 \text{ is the optical admittance of the substrate and incident medium and } " * " \text{ denotes the complex conjugate operation. The measurand } I_r \text{ can therefore be calculated based on Eq.(1) and Eq.(3). To achieve better sensing performance, } I_r \text{ should be adjusted to the minimal value when no external loading so that the monotonic sensing range can be maximized. Also, since the optical response is sinusoidal nature, the Fabry-Perot microcavity should be optimized to minimize nonlinearity throughout the sensing range. The optimal microcavity is shown in Fig. 5 and the final output response of the proposed dual optical fiber optical pressure sensor is shown in Fig. 6. As shown in Fig.6, the percentage nonlinearity of the proposed optimal sensor is successfully limited within 2\%, which is evidently better than that of the optimal pressure sensor based on single optical fiber detection scheme [11].}

5. Fabrication process

In the previous work, we have already successful fabricated the proposed Fabry-Perot pressure sensor without packaging sleeves [6]. The major fabrication process is shown in the Fig.7.
Silicon nitride layers are first formed by low-pressure chemical vapour deposition (LPCVD) on both sides of the silicon wafer and are used as the mask for subsequent KOH etching to form a single deep cavity as shown in Fig. 7(b). Afterwards, the fixed mirror layer, sacrificial layer (PSG) and moving mirror layer are deposited by LPCVD successively as shown in Fig. 7(c). Etching windows are then opened by RIE. Finally, diaphragm is released in 40% HF solution as shown in Fig. 7(d). Therefore, it can be seen that in order to fabricate the proposed packaging sleeve to hold the optical fibers, biocompatible SU-8 layer should be formed before the final release process of the diaphragm.

6. Conclusions

We proposed a novel SDCD based dual optical fiber Fabry-Perot pressure sensor for blood pressure measurement. Compared with conventional intensity-modulated pressure sensor using single optical fiber, such type of sensor can increase the measurement linearity and avoid the phantom pressure measurement induced by optical intensity loss. The mechanical and optical simulation results show that the proposed pressure sensor can successfully measure the blood pressure range from 0 to 150 mmHg with less than 2% nonlinearity. Also, the fabrication process of the proposed sensor is given, instead of complicated fusion bonding process, only bulk and surface micromachining techniques are required which facilitate the mass production of such biocompatible and disposable pressure sensors.

References
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