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Microfiber-based Intraocular Pressure Sensors: a Proposal

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Abstract: In this paper, we designed a microfiber-based intraocular pressure sensor that incorporates an optical microcavity in the fiber covered by a flexible PDMS film. The whole sensor head is about 10 microns in diameter and 6 microns in thickness. The result shows that the deformation of the film and optical response with respect to externally applied pressure is linear, showing the desired properties. Though there is a long way to experimentally demonstrate the sensor, we believe this proposed structure could help in designing more efficient and compact intraocular pressure sensors with remote read-out.
Keywords: microfiber-based intraocular pressure sensors, optical microcavity, fiber, flexible PDMS film.

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1. INTRODUCTION

Diagnosing and monitoring the development of glaucoma involves taking regular measurements of patients' intraocular pressure [1-2]. This is typically done in a clinical setting with readings taken weeks or months apart, a problem since the intraocular pressure can rise and fall unexpectedly throughout the day and may be subject to the white coat effect [3-4]. The current consensus among ophthalmologists and optometrists define normal intraocular pressure as that between 10 mmHg and 20 mmHg. The average value of intraocular pressure is 15.5 mmHg with fluctuations of about 2.75 mmHg [1].

Here, we propose a highly miniaturized intraocular pressure sensor using a microfiber-based optical cavity. The front surface of the cavity is deformable to external pressure, leading to the interference change of the spectrum of the optical signal inside the cavity. The proposed sensor has the advantage of being immune to electromagnetic interference, compact size and possess a linear response.

Figure 1. A patient in front of a tonometer.

Typically, tonometry is performed using a relatively bulky apparatus, greatly decreasing its usability and availability to patients [1].

2. THEORETICAL BACKGROUND

In this model, we need to study the mechanical response of the device first. In an arbitrary Cartesian coordinate system, the force and displacement vectors can be represented by 3 × 1 matrices of real numbers. Then the tensor κ connecting them can be represented by a 3 × 3 matrix of real coefficients. When multiplying the tensor by the displacement vector, one gets the force vector [5]:

$$\mathbf{F} = \begin{bmatrix} F_1 \\ F_2 \\ F_3 \end{bmatrix} = \begin{bmatrix} \kappa_{11} & \kappa_{12} & \kappa_{13} \\ \kappa_{21} & \kappa_{22} & \kappa_{23} \\ \kappa_{31} & \kappa_{32} & \kappa_{33} \end{bmatrix} \begin{bmatrix} X_1 \\ X_2 \\ X_3 \end{bmatrix} = \boldsymbol{\kappa}\mathbf{X} \qquad (1)$$

where F and X is the force and displacement vector while κ ij (i, j = 1, 2, 3) is the spring constant tensor.

In order to evaluate the optical performance of the device, we numerically calculate the reflected signal due to the interference by the following equation [6].

$$\text{Reflection} = \frac{4R\sin^2\delta}{(1-R)^2 + 4R\sin^2\delta} \qquad (2)$$

where R is the reflection of each surface and $\delta = 2\pi L/\lambda$ is the optical phase introduced in each path of the light inside the cavity. L is the single path distance.

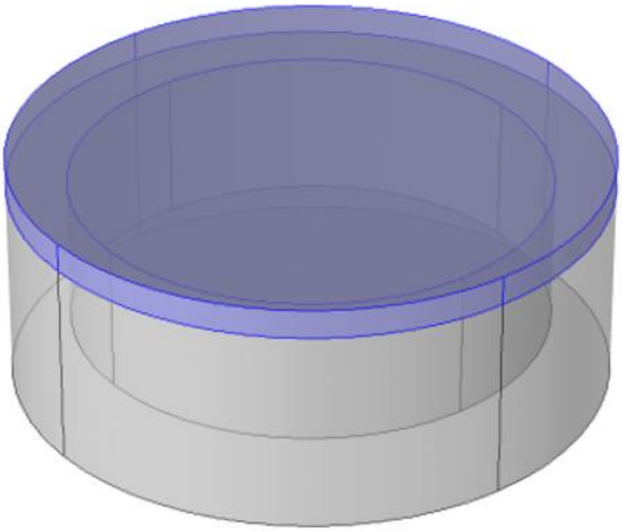


Figure 2. Schematic of the pressure sensor studied.

The semi-transparent blue part is the PDMS film with thickness of 1 um. The film is sealed to the glass fiber with an air cavity inside. The radius of the microfiber is 10 um while the cavity is 8 um.

3. MODEL DEFINITION

The model consists of a microfiber head etched to a cylinder cavity (grey part) and a PDMS cover (blue part) as a deformable reflective surface. The illustrative picture shown below. The cavity is 8 um in radius and 5 um in thickness, considering both miniaturization requirement and real fabrication demands. The top of the fiber cavity is sealed with PDMS, a common polymer used in bio-applications. PDMS has a suitable mechanical property (Young’s modulus = 800 [kPa] and Poisson’s ratio = 0.45 as listed in Table 1) in this particular situation. It deforms easily under external pressure. All other parameters used in the simulation are presented in Table 1. During simulation, a parameter sweep is performed with the external pressure ranging from 0 mmHg to 26 mmHg and deformation displacement is extracted at the center of the PDMS film. The displacement is further used for simulating the optical signal of the interference. The reflection is assumed to be 80% considering that the surface of the two surfaces can easily be coated with metal films. The high reflectivity will help increase the signal-noise ratio in determining the resonant wavelength.

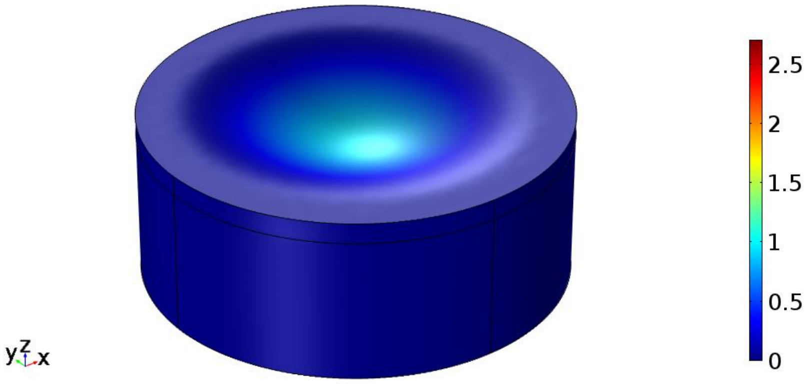
Table 1. Parameters used in the model to define geometry and materials’ parameters.

P_input	20[mmHg]	external applied pressure
r_fiber	10[um]	radius of microfiber
r_gap	8[um]	radius of the air gap
d_gap	5[um]	gap of the air
d_film	1[um]	thickness of the film

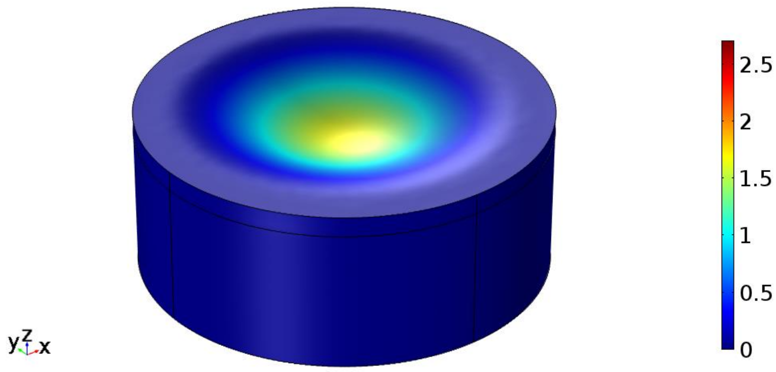
L_fiber	8[um]	length of the fiber
rho_PDMS	970[kg/m^3]	density of PDMS
rho_SiO2	2220[kg/m^3]	density of SiO2
E_PDMS	800[kPa]	Young's modulus of PDMS
E_SiO2	73[GPa]	Young's modulus of SiO2
nu_PDMS	0.45	Poisson's ratio of PDMS
nu_SiO2	0.167	Poisson's ratio of SiO2

4. RESULTS AND DISCUSSIONS

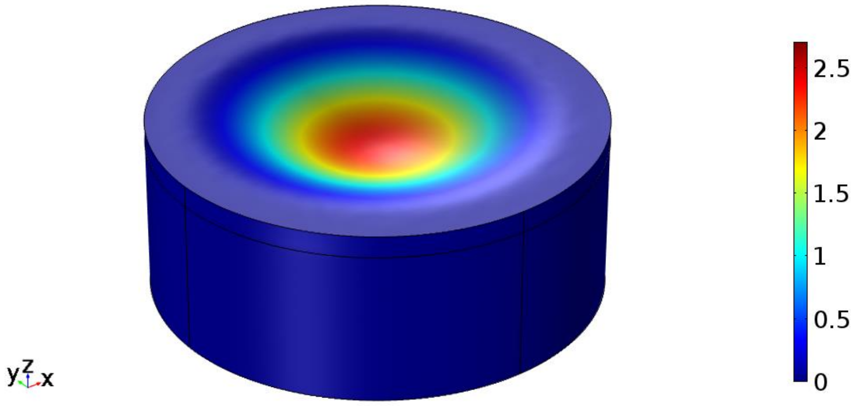
Figure 3 shows the displacement distribution device under different pressure: 8, 14, 20 and 26 mmHg. It is easily seen that the displacement increases with the applied pressure. The deformation occurs mainly in the center of the film within a diameter of around 4 microns which is comparable to the typical spot size of the optical wave propagating inside [7]. The displacement at the center of the film with respect to the pressure is plotted in Figure 4. It is easy to find that the displacement increases linearly from 0 to 3.2 um. The linearity is important as it will facilitate our determination of the resonant spectrum more precisely.



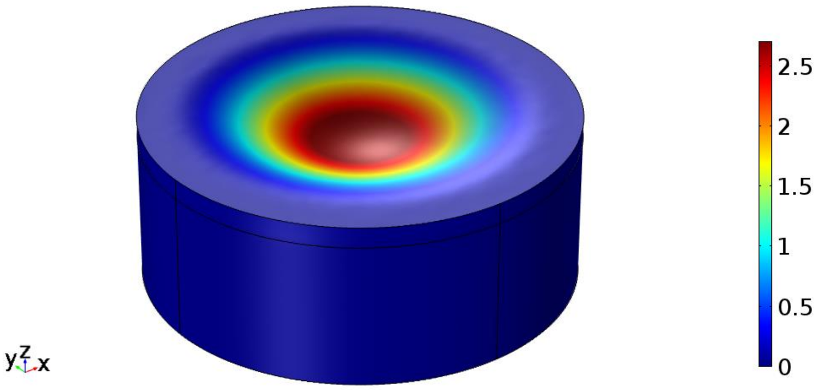
(a)



(b)



(c)



(d)

Figure 3. Displacement distribution of the device under different pressure: 8, 14, 20 and 26 mmHg from top to bottom.

We further calculated the optical spectrum of the cavity formed by the fiber and the PDMS film by employing Equation (2). The Matlab code is included in the supplementary part at the end of the paper. As the displacement changes, the phase (δ) that the light acquires changes accordingly, leading to the shift of the spectrum. The spectrum is calculated from 1.2 um to 2.4 um in wavelength, corresponding to the operation wavelength of conventional single mode fibers. It is clear that the resonant wavelength of the device differs under different pressure. Here, we want to draw the readers' attention to the interesting spectrum centered around 1.3 um due to the reasons following. First, the resonant dip of the spectrum shifts to a shorter wavelength as the pressure increases. This is reasonable because when the pressure increase, the displacement increase, and the actual cavity length decreases, resulting in blue-shift. Second, the shift is monotonic. That means when we monitor the resonant dip to determine the pressure, we will not come to a difficulty in tracking the pressure. Third, the shift is quite linear, especially from 10 mmHg to 20 mmHg (data not shown), corresponding to the eyes' normal working range.

Here, we want to point out that the design shown here is not the best. We have reserved rooms for improvement, such as optimizing the thickness of the film, increasing the reflectance of each surface for better performance, choosing better materials for real applications, etc.

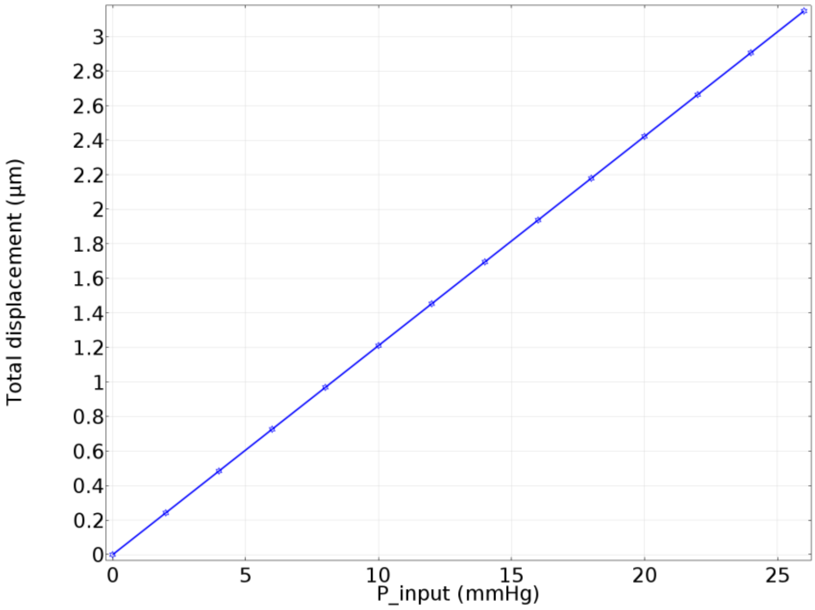


Figure 4. Displacement at the center of the PDMS film as a function of the external applied pressure. Note that the response of the film to the pressure is quite linear.

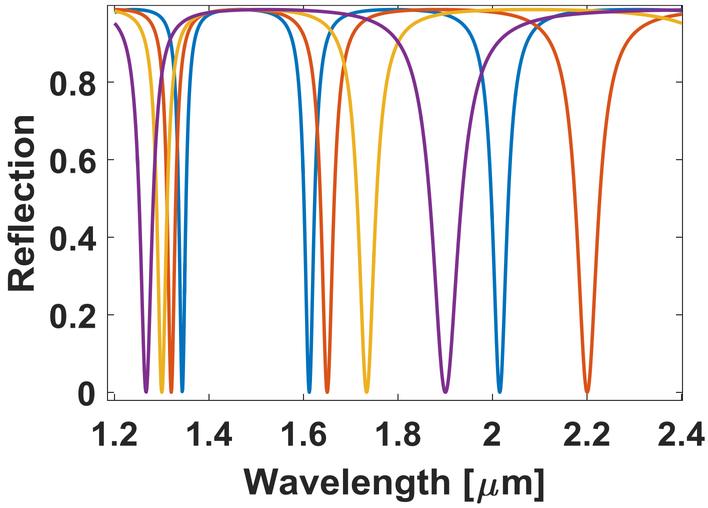


Figure 5. Optical interference spectrum of the device at external applied pressure of 8 (blue), 14 (red), 20 (orange) and 26 mmHg (purple), respectively.

5. SUPPLEMENTARY INFORMATION

```
clear;
R=0.8; % reflectance of each surface
d_gap=5e-6; % gap distance of the cavity
for displacement= [0.97 1.7 2.4 3.1] *1e-6;
    lambda= [1.2:0.001:2.4] *1e-6; % wavelength
    delta=2*pi*(d_gap-displacement). /lambda; % phase
    Reflection=4*R*(sin(delta)). ^2. /((1-R) ^2+4*R*(sin(delta)). ^2);
    plot (lambda*1e6, Reflection,'linewidth',2);
    hold on;
```

```
end
xlabel('Wavelength [\mum]');
ylabel('Reflection');
```

6. CONCLUSION

In conclusion, we have designed an intraocular pressure sensor based on microfibers. The sensor is characterized in details regarding external pressure stimulation. The device is compact in size and is in linear in mechanical and optical response, the desired characteristic for sensors. In addition, it only requires common commercial materials found in the market making it potentially affordable. Though aspects regarding the device remain elusive, it shows potentials for demonstration.

ACKNOWLEDGEMENTS

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REFERENCES

[1] https://en.wikipedia.org/wiki/Intraocular_pressure.
[2] www.glaucoma.org (Glaucoma Research Foundation).
[3] J. Danias, A. I. Kontiola, T. Filippopoulos, and T. Mittag, “Method for the Noninvasive Measurement of Intraocular Pressure in Mice,” IOVS, vol. 44, no. 3, pp. 1138-1141, Mar. 2003.
[4] H. Bagga, J. H. Liu, and R. N. Weinreb, “Intraocular pressure measurements throughout the 24h,” Current Opinion in Ophthalmology, vol. 20, no. 2, pp. 79-83, Mar. 2009.
[5] Alberto Carpinteri, Structural Mechanics Fundamentals, CRC Press, 2013.
[6] J. D. Jackson, Classical Electrodynamics, 3rd edition, Wiley, 1999.
[7] Joseph C. Palais, Fiber Optic Communications, 5th edition, Prentice Hall, 2004.