A white-light source operated polymer-based micromachined Fabry-Perot chemo/biosensor

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Abstract — A white-light source operated polymer-based micromachined Fabry-Perot chemo/biosensor is reported for the first time. It is a refractive-index sensitive optical sensor. The transducing signal varies upon the changes of the effective index of refraction in the sensing area, which is the Fabry-Perot cavity. This chemo/biosensor is made with PDMS and glass. Specifically, the micromachined Fabry-Perot interferometer (µFPI) and cavity is fabricated by bonding glass and the soft-lithographically patterned PDMS. Several chemicals have been detected with the prototype devices. Measurements show that at least 10ppm chemicals can be detected even without any performance optimization of the devices.

Keywords — chemosensor, biosensor, polymer Fabry-Perot interferometer, white light source

I. INTRODUCTION

Optical technique is very attractive for sensing applications due to its non-invasive nature, high degree of sensitivity, capability of multiplexing and immunity to environment noise. Many optical components or systems such as lenses, gratings, mirrors, micro ring resonators and interferometers have been miniaturized as various sensors for different applications [1-5]. For instance, micromachined Fabry-Perot interferometers (µFPIs) have been designed and implemented for chemical sensing, gas sensing, ultrasonic sensing and as optical modulators [5-13]. For chemical and gas sensing, the FP cavity serves as sensing area. The output signals are transmitted intensity through FPI, which vary with different chemicals [7] or gases [9]. For ultrasonic sensing, the micromachined capacitive acoustic transducer consists of a FPI cavity with embedded optical diffraction gratings on a transparent substrate. The detection sensitivity can be maintained at an optimum level by deflecting the membrane of the FPI [6]. For optical modulators, a FPI contains a crosslinked electro-optic polymer inside its cavity, offering the potential for high time-bandwidth modulation [5]. Various MEMS actuation mechanisms enabled a lot of possible unique features for a µFPI. For example, actuation makes tunable µFPIs possible. Tunable µFPIs have been developed for wavelength division multiplexing (WDM) in optical communications [8] and Raman spectroscopy [11], oxygen detection in blood sample [12] and spectral endoscope optical imaging [13]. For WDM application, tunable µFPI serves as a tunable filter, selecting different wavelengths by changing the gap size of the FPI cavity. For oxygen detection in blood sample, this proposed µFPI device can be used in a wide wavelength range from visible to near infrared light, offering sufficient characteristics to analyze the spectrum of the blood [12]. For spectral endoscope optical imaging, the device can acquire spectral images of a target at each pixel [13]. Recently, µFPI has been also used to study the nanoscale fluid dynamics [14], indicating its potential for nanoscience and nanotechnology research.

Usually µFPIs are fabricated from silicon, polysilicon, silicon nitride, silicon oxide thin film or other semiconductor materials and are often operated using a laser source. Herein a polymer-based µFPI chemo/biosensor operated with white-light source is reported for the first time.

II. DEVICE DESCRIPTION AND MODELING

The schematic and operational principle of the white-light source operated µFPI chemo/biosensor is given in Fig. 1 and Fig. 2. It consists of a PDMS plate and a glass plate. A µFPI cavity with a gap of 50 µm is formed between them, serving as the sensing area. At current stage, both plates do not have any thin film coating to enhance their reflectivity like traditional macroscale FPI or reported µFPI.

The operation light undergoes multiple internal reflections upon entering the cavity between the two plates and interferences inside the cavity, resulting in modulated output transducing signals (e.g., transmitted and reflected interference fringes) as show in Fig.1 and Fig. 2. Specifically, the operation principle of this sensor is as the following: If the effective index of refraction inside the FP cavity changes due to the presence of different chemicals or the binding between antibodies (Abs) and antigens (Ags), the interference fringes shift. Its performance is determined by its finesse $F$, which is related to the free spectra range (FSR=$\Delta \lambda$) and the full-width at half-maximum (FWHM)$\delta \lambda$ [15].

![Fig. 1(left) Schematic of a PDMS-based µFPI chemo/biosensor, showing immobilized antibodies (Abs) on the surface of the Au-coated glass surface and the binding between Abs and antigens (Ags); (right) operational principle of the FPI, showing the reflected and transmitted light through the FPI as output transducing optical signals.](image-url)
Fig. 2 Output transducing signal (transmitted or reflected signal) from white-light source operated FPI [15]

The fiber-optics based testing setup is illustrated in Fig. 3. The custom-designed optical fiber probe delivers the white light to the sensor and also receives the reflected transducing signals from the µFPI sensor, which are coupled to a spectrometer.

Fig. 3 Optical setup: the optical fiber probes consist of a tight bundle of 7 optical fibers in a stainless steel ferrule. The center fiber is to collect the reflected light while the outer 6 fibers deliver illumination light to the FPI chemo/biosensor;

The sketch of the optical system including the optical fiber and the µFPI for the modeling is illustrated in Fig. 4. The white light delivered to the sensor can be modeled as a Gaussian beam [8]. The loss of the optical coupling between the fiber and µFPI sensor needs to be considered.

Assume the loss coefficient of µFPI cavity is $L$ and the loss between the fiber probe and µFPI is $L_{fiber-FPI}$, then the reflected intensity $I_r$ from this system, coupled to the spectrometer, is written as:

$$I_r = I_i \exp(-\frac{2(\lambda - \lambda_0)^2}{\omega^2}) \times L_{fiber-FPI} \times f(R_{air-PDMS}, R_{air-glass})$$  \hspace{1cm} (1)

Where

$$f(R_{air-PDMS}, R_{air-glass}) = \frac{r_2^2 + r_3^2 - 2r_2r_3L\cos(2n_2d)}{1 - r_2^2 + r_3^2 - 2r_2r_3L\cos(2n_2d)}$$

and

$$r_2 = -r_1 = \frac{n_2 - n_1}{n_2 + n_1}$$

$$r_3 = \frac{n_2 - n_3}{n_2 + n_3}$$

Where $I_i \exp(-2(\lambda - \lambda_0)^2/\omega^2)$ is incident light intensity; $f(R_{air-PDMS}, R_{air-glass})$ is the reflectivity from the µFPI; $R_{air-PDMS}$, $R_{air-glass}$ are the reflectivity at the interface between the air and PDMS, and air and glass, respectively and $d$ is the gap size of FPI cavity.

III. DEVICE FABRICATION

The device was fabricated using an inexpensive and rapid soft lithography process (Fig. 5). Briefly, a 50 µm thick SU8 mold of the device is formed on a silicon substrate using optical lithography. PDMS is casted on the mold, followed by 1.5 hour curing at the temperature of 65 ºC. The PDMS layer and the glass substrate are bonded together to complete the device fabrication after oxygen plasma treatment. The input and output wells are made in the PDMS layer thereafter.

Fig. 5 The fabrication process flow: (a) form the 50µm tall SU8 mold on the silicon wafer; (b) transfer the mold to PDMS; (c) bond patterned PDMS to glass wafer after plasma treatment, and then form the input/output wells

An optical micrograph of the SU8 mold of one single µFPI
device showing the integrated channel for samples delivery to the FP cavity is given in Fig. 6. The diameter of the FPI plate is 250 µm. The cavity length is 50 µm. Optical images of arrayed PDMS FPI devices are shown in Fig. 7.

IV. MEASUREMENTS AND DISCUSSIONS

The optical source for the experiments is a tungsten halogen light source (Ocean Optics Inc.), which is a versatile white-light source optimized for the VIS-NIR (360-2500 nm).

The measured output signal from a µFPI device with air inside its cavity is given in Fig. 8. Based on the model proposed in section II, the calculated transducing signals with air in the FPI cavity match the measured signals quite well as shown in Fig. 8. Note that the intensity of the white light source across the wavelength range from 360-2500 nm is not uniform. The envelope of the interference fringes is the profile of the intensity of the white light source.

This prototype sensor has been evaluated with chemicals such as ethanol, DMEM-electrolyte and water. Clear shift of the interference fringes has been observed as shown in Fig. 9. Theoretically, the detection of limit (DOL) of the sensor is only limited by the resolution of the spectrometer. In order to evaluate the DOL of the sensor, several ethanol samples with different concentrations have been tested. Measurements found that the DOL of the ethanol can be at least lower than 10ppm for this prototype sensor without any performance optimization (Fig.10). Another important performance parameter of the sensor is its resolution, the minimum change of the refractive index inside the µFPI cavity which a µFPI can distinguish. More experiments are underway to investigate the resolving capability of this sensor.

In order to enhance the DOL of this sensor, the transducing optical signals need to be amplified, especially if the amount of the sample to be tested is tiny. Specifically designed plasmonic nanostructures may fulfill this requirement, resulting from the localized surface Plasmon resonance effect.
For its resolution improvement, the finesse $F$ of the µFPI needs to be improved [15]. Assuming the reflectivity of the mirror is $R$, its finesse is given by $F = \pi \sqrt{R} / (1 - R)$. Maximization of finesse requires bringing $R$ as close as possible to 1. This can be realized by properly designed thin film coating. Furthermore, for traditional µFPI, the sensing area is limited by the area of the size of the µFPI mirror plate. Some specific integrated nanostructures on the mirror plate can increase the sensing area up to two orders of magnitude larger than that of a traditional µFPI, leading to further improvement of its sensitivity. This technique is under development in our lab.

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REFERENCES