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A refractive-index biosensor based on directional coupling of silica waveguides

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ABSTRACT

We propose a refractive-index biosensor based on directional coupler (DC) structure to detect physiological concentrations of glucose in water by sweeping of incident light wavelength. The sensing mechanism is by changing the incident light wavelength the refractive index change induced by glucose concentration can be compensated to keep the coupling length of DC unchanged. The influences of the central light wavelength, the waveguide width and spacing on the sensitivity and resolution of the proposed sensor are analyzed. By a tradeoff between the coupling length and resolution in a sample index range of 0.02, we can obtain a sensor with a size of 33.65 mm × 100 μm for $d_1 = d_2 = 3$ μm which shows a resolution of about 1.8×10^{-5} and a sensitivity of 5500 nm/RIU using a tunable laser with a precision of 0.1 nm.

1. Introduction

Biosensing attracts growing interest in scientific research as its wide variety of applications in medical and environmental applications because of their small size, chemical and biological compatibility and electromagnetic insusceptibility [1–3]. Biosensors hold great promise to develop fast, inexpensive, portable biomedical devices for point-of-care diagnostics and healthcare applications [4–8]. For an ideal biosensor, it must show a very high sensitivity, quick response, small size, portability and low cost.

As modern biosensors should be highly selective and sensitive as well as rather compact, many optical methods have been developed to meet these demands, such as optical fiber [9], directional coupler (DC) [10], Mach–Zehnder interferometry [8], ring resonators [11], and surface plasmon resonance (SPR) [12]. Among the various techniques, DC has been one of the widely useful methods in the development of biochemical sensors in recent years. Optical biosensors can basically be considered as an optical transmission line between a source and a detector. A bio-optical transduction mechanism generally changes power, spectrum, polarization or the delay of the transmitted optical signal according to the properties of the investigated analyte medium which is typically located inside of a sealed or permeable cell. We have studied the case that the refractive index change brings the change of output power [10]. The refractive index sensor has a sensing resolution of 2.25×10^{-5} for index change at around 1.455, while the sensitivity is about 13.36/RIU which is needed to be improved.

In this paper, we propose a sweeping wavelength method in a silica waveguide with DC structure. The refractive index change induced by glucose concentration is compensated by the dispersion of silica waveguide to keep the coupling length of DC unchanged. In this method the refractive index can be detected by peak wavelength change and therefore the glucose concentration is obtained. The sensitivity and resolution of the refractive index sensor are studies for different configuration of DC.

2. Refractive index sensing mechanism and theoretical analysis

In a refractive index sensor with a DC structure as shown in Fig. 1, the silica (n_1) waveguide is surrounded by the glucose water sample (n_g) to be measured in the sensing area. Here we make use of d_1 and d_2 to denote the width of silica waveguides and the separation distance between them. We set the coupling length of the DC as that with water ($n_0 = 1.333$) in sensing area and in this case the incident light with a wavelength of λ_0 is totally coupled into Port 2, which means P_{out1} is zero and P_{out2} is maximum.

As a sample with different glucose concentration brings a change of coupling length in DC, the output of Port 1 (P_{out1}) is nonzero. By sweeping the incident light wavelength, we can get

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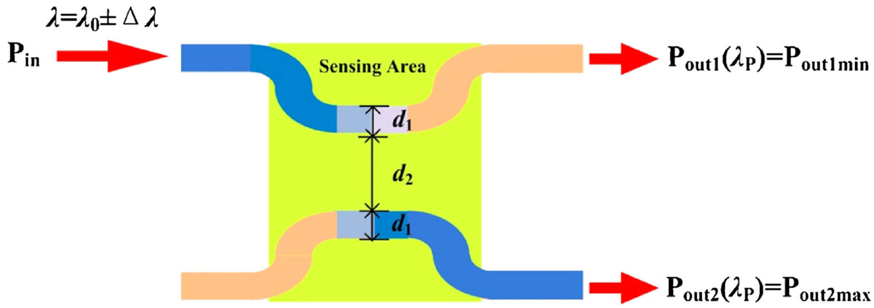

Fig. 1. The top-view of biosensor with a DC structure.

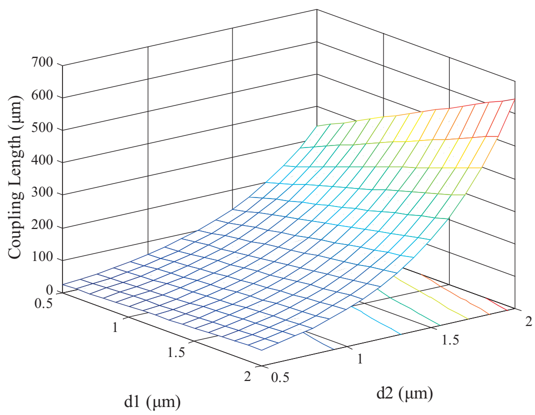

Fig. 2. Coupling length of the DC sensor with different waveguide width and spacing.

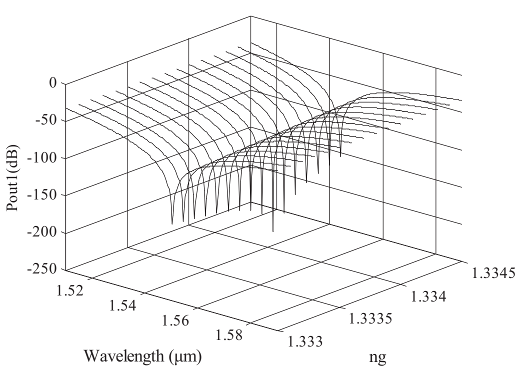

Fig. 3. The output of Port 1 for different wavelength and sample index.

a minimum P_{out1} and a maximum P_{out2} for a special wavelength λ_p, which is termed as the peak wavelength. This principle is based on the dispersion relation of silica waveguide [13]:

$$n_1^2 = \frac{0.6961663\lambda^2}{\lambda^2 - 0.0684043^2} + \frac{0.4079426\lambda^2}{\lambda^2 - 0.1162414^2} + \frac{0.8974794\lambda^2}{\lambda^2 - 9.896161^2} + 1 \tag{1}$$

The coupling wavelength change can be compensated by the refractive index change for a different light wavelength (λ_p) to keep the coupling wavelength unchanged. Therefore we can build a relation between the glucose concentration and peak wavelength, and the refractive index sensor is realized.

The coupling effect of the DC can be analyzed by interference phenomena between the even mode and odd mode, and the electric field in the directional coupler can be approximated by the summation of even mode and odd mode when high-order modes are neglected. Here we make use of β_e and β_o to denote the propagation constant of the even mode and odd mode, respectively, and

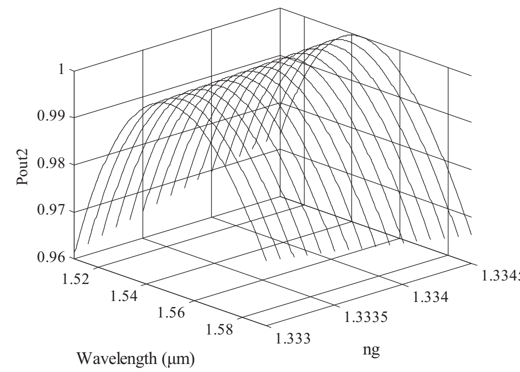

Fig. 4. The output of Port 2 for different wavelength and sample index.

they can be obtained by solving the following equations

$$2u = a\tan\left(\frac{n_1^2 w}{n_0^2 u}\right) + a\ \tan\left[\frac{n_1^2 w}{n_0^2 u}\tanh\left(\frac{d_2}{d_1}w\right)\right] \quad \text{(even mode)} \tag{2}$$

$$2u = a\tan\left(\frac{n_1^2 w}{n_0^2 u}\right) + a\ \tan\left[\frac{n_1^2 w}{n_0^2 u}\coth\left(\frac{d_2}{d_1}w\right)\right] \quad \text{(odd mode)} \tag{3}$$

where $u = \frac{d_1}{2}\sqrt{n_1^2 k_0^2 - \beta^2}$ and $w = \frac{d_1}{2}\sqrt{\beta^2 - n_0^2 k_0^2}$.

The coupling length of the five-layer waveguide can be written as

$$L_c = \frac{\pi}{\beta_e - \beta_o} \tag{4}$$

The corresponding mode-coupling coefficient is also obtained

$$\kappa = \frac{\pi}{2L_c} = \frac{\beta_e - \beta_o}{2} \tag{5}$$

For simplicity, we assume each of β_e and β_o equals the summation of β_0 and a different perturbation, where β_0 is defined as the propagation constant of the single-slab waveguide and can be calculated by solving the equation

$$u_0 = a\tan\left(\frac{w_0}{u_0}\right) \tag{6}$$

And the coupling coefficient can be obtained as

$$\kappa = \frac{4u_0^2 w_0^2}{\beta_0 d_1^2 (1 + w_0^2) v_0^2} \exp\left(\frac{2d_2}{d_1} w_0\right) \tag{7}$$

where $v_0^2 = \frac{1}{4} k_0^2 d_1^2 (n_1^2 - n_0^2)$. The coupling length is

$$L_c = \frac{\pi}{2\kappa} = \frac{\beta_0 d_1^2 (1 + w_0^2) v_0^2 \pi}{8 u_0^2 w_0^2} \exp\left(-\frac{2d_2}{d_1} w_0\right) \tag{8}$$

We give the coupling length for different d_1 and d_2 ranging from 0.5 μm to 2 μm as shown in Fig. 2. When the waveguide width