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Label-free optical biosensor based on a dual-core microstructured polymer optical fiber

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ABSTRACT

A design for high-sensitivity and label-free optical biosensor based on the principle of wavelength-selective resonant coupling between the cores of a dual-core microstructured polymer optical fiber (MPOF) is demonstrated. An immobilized antigen sensing biolayer on the walls of the holes in a water-filled microstructured core provides the ability to selectively capture antibody biomolecules. The change of the thickness of antibody biolayer can be detected based on the shift of the resonant wavelength, which is achieved between a water-filled microstructured core and a small core with a small air hole in the center through selectively directional resonant-coupling. Accordingly, the transmission spectra of the output port will present a notch at the resonant wavelength, which is very sensitively dependent on the thickness of the biolayer. Numerical simulations demonstrate that the proposed biosensor can get a sensitivity of up to 5.6 nm/nm. In addition, the sensor resolutions of the biolayer thickness as high as 0.012–0.015 nm are presented in the whole 600–840 nm region.

1. Introduction

Microstructured optical fibers (MOFs) have been successfully applied to the sensing fields, as their holes can be controllably filled with ultrasmall volumes of analytes [1–4]. In particular, microstructured optical fibers made of silica for refractive index sensing purposes has been well explored, which are based on a Bragg or long period grating (LPG) [5–7], surface plasmon resonances [8–10], photonic bandgap properties of the PCF [11,12], or resonant coupling [13–20]. In addition, MOF based sensors have achieved considerable attention especially in the context of label-free fiber-optic biosensing, in which biomolecules does not require the sample to be marked with fluorescent dyes but rather relies on the detection of tiny refractive index changes due to bonding events, e.g., with antibodies or antigens. Rindorf et al. experimentally demonstrated a sensitivity of 1.4 nm shift of a long-period grating resonance per nm biolayer (1.4 nm/nm) [7], and Ott et al. [21] theoretically demonstrated a sensitivity of 10.4 nm/nm using the inherent nonlinearity of the photonic crystal fiber (PCF) for Four-Wave-Mixing (FWM)-based label-free biosensing, and thus an experimental work on using the inherent material nonlinearity of the fiber in a FWM sensing scheme has been investigated [22]. Furthermore, a biosensor based on anti-resonant reflecting optical waveguide has been theoretically demonstrated in the optical region for detection of the deposited monolayer on the hollow-core surface of fibers [23]. Recently, an application of metallized photonic crystal fibers in surface plasmon resonance sensors of biolayer thickness is demonstrated, where sensor resolution of the biolayer thickness is as high as 0.039–0.044 nm in the whole 600–920 nm region [9]. And very recently, You et al. [24] experimentally demonstrated a simple dielectric hollow-tube at terahertz range for bio-molecular layer sensing based on the anti-resonant reflecting wave-guidance mechanism. In addition, MOFs can also be used as the sensing element in labs-on-a-chip [25].

In fact, the above-mentioned biosensor can also be applied to detecting the refractive index of materials. In particular, refractive index sensors taking advantage of coupling characteristic between dual-core MOFs have been proposed recently [16–20]. Town et al. presented a dual-core microstructured optical fiber with a sensitivity of up to 169,711% by using the exponential dependence of intercore coupling on the refractive index of the analyte [18]. Yuan et al. demonstrated a sensitivity of 70,000 nm/RIU in an all-solid dual-core photonic bandgap fiber, where a single hole between the cores acts as a channel for the analyte [19]. However, the background of the above reported dual-core is almost based on silica. In the last few years, MOF based on polymers, such as PMMA [26] or topas [27–29] are paying an important role in optical biosensing. We theoretically describe a dual-core MPOF biosensor made

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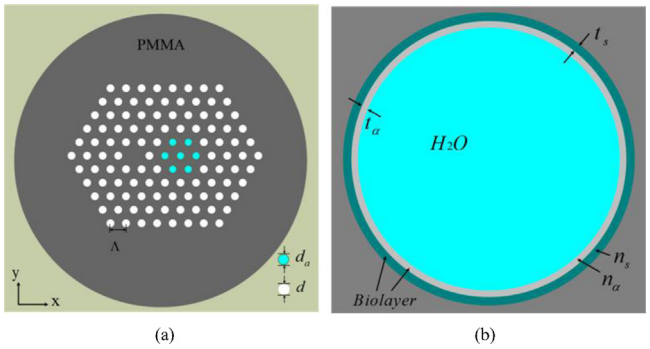

Fig. 1. (a) Cross-section of a dual-core MPOF biosensor with pitch $\Lambda = 2.8\,\mu m$ and hole diameter $d = 1.3\,\mu m$. (b) Water filled hole with a sensor layer and an attached layer of biomolecules.

of poly(methyl methacrylate) (PMMA), which is composed of two asymmetric cores where one core is formed by introducing a small air hole into the center of a solid core, while the other is an analyte-filled microstructured core. The fiber biosensor proposed here can be fabricated by two steps. Firstly, a MPOF with a small air-hole in one core is fabricated by the technique reported by Barton et al. [30] and Cordeiro et al. [31]. Secondly, the seven air-holes are selectively filled with the anaylte by the similar technique proposed previously [31–34] and the whole processing are performed with great care in order to minimize errors from coupling instabilities, cleaving quality, bending effects, etc. [35]. We know that the seven analyte-filled air-holes forms a microstructured core. In fact, making multi-holes in the core of an index-guided MOF is a new way to help to increase the amount of energy that travels in the holes where the analyte can be filled [31]. We can see that the cores of the MPOF form an asymmetric directional coupler. Like silica MOF couplers, the operation of couplers in microstructured polymer optical fibers (MPOFs) typically involves energy transfer over a certain coupling length between the two fiber cores coupled by proximate interactions. Specifically, at the phase matching wavelength, the maximum power coupling from one core to the other occurs. As a result, the spectral transmission of the input core presents a notch corresponding to the index-matched wavelength, when a broadband beam is launched into one core. The notch is sensitively dependent on the thickness of the biolayer. Numerical simulations show that the biosensor presents a sensitivity of around 5.6 nm/nm and sensor resolutions of the captured antibody thickness as high as 0.012 nm are demonstrated in the whole 600–840 nm range.

2. Numerical simulation

An initial dual-core MOF made of PMMA in a hexagonal hole index-guiding structure with $\Lambda = 2.8\,\mu m$ and hole size $d = d_a = 1.3\,\mu m$ is shown in Fig. 1(a). The main advantage of considering PMMA is that biomolecules can be attached directly to the surface of the holes of the fiber [36], avoiding in this way any further functionalization. Recently, Markos et al. presents feasible designs of a dual-core MPOF for detecting the thickness of biomolecules and achieves a highly sensitive and selective biosensor [37]. The procedure for capturing α-streptavidin is discussed by Jensen et al. [36], where α-streptavidin antibody was captured by an antigen-antibody interaction with streptavidin. Owing to the fact the streptavidin molecules can bind directly to the polymer surface while still being able to bind antibody, we consider that inside the holes a thickness of the antigen layer of t_s has been immobilized onto the walls. Then, this sensor layer can capture a layer of antibody biomolecules of thickness t_α through a selective biochemical process, as shown in Fig. 1(b). It can be seen from Fig. 1(b) that the holes of the microstructured core are filled with water. This is due

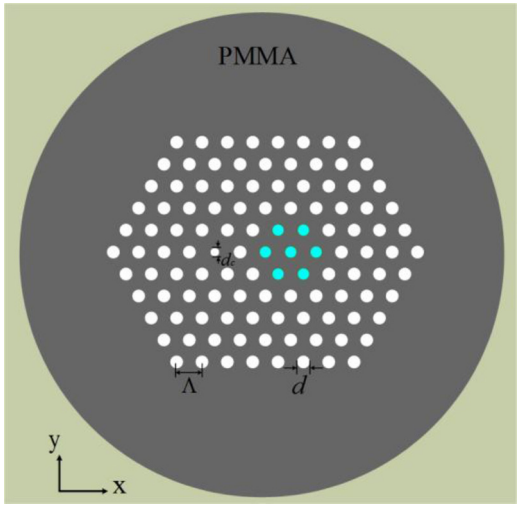

Fig. 2. Cross-section of a dual-core MPOF biosensor with pitch $\Lambda = 2.8\,\mu m$, hole diameters $d_a = d = 1.3\,\mu m$ and small air hole diameter $d_s = 0.84\,\mu m$. In addition, water filled hole with a sensor layer of $t_s = 40$ nm and an attached layer of biomolecules of thickness $t_\alpha = 5$ nm.

to the fact that the introduction of samples in aqueous solution into the sensor might leave remnants of water, which from an experimental point of view is difficult to remove [37]. Here we assumed that the refractive indices of the biolayers have the same refractive index of 1.45 and neglected dispersion of the biolayers.

In fact, it should be noted that the phase matching is difficult to realize for the configuration shown in Fig. 1(a) owing to the obviously large mode indices difference and the index difference between the two cores is always very large. Therefore, a small hole of diameter d_s into the fiber core center assures lowering the effective refractive index of the core guiding mode (in order to facilitate phase matching with the water-filled microstructured core) is proposed and the cross-section of which is depicted in Fig. 2. Then, maximum power transfer over a coupling length occurs at an index-matched wavelength, namely a resonant wavelength. As thus, the spectral transmission of the left core presents a notch corresponding to the resonant wavelength, where a broadband beam is launched into the left core.

We solve the modes of the individual cores in the proposed biosensor (see Fig. 2) by semi-vectorial beam propagation method (BPM) [38]. Fig. 3 plots the effective indices of the fundamental modes in the two cores of the dual-core MPOF shown in Fig. 2. As is shown in Fig. 3, when no sensor layer is attached to the side of the holes, the phase matching cannot be realized owing to the

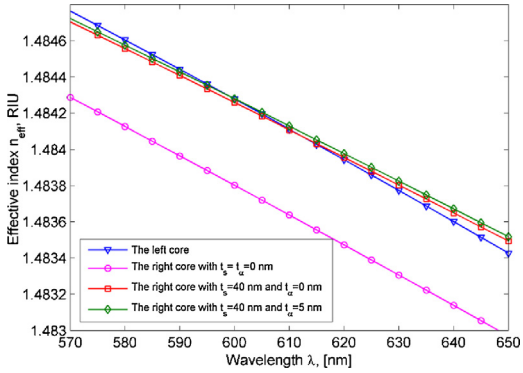

Fig. 3. The effective index curves of the fundamental modes between the left core and right core for the vertical polarization with $\Lambda = 2.8\,\mu m$, $d_s = 0.84\,\mu m$, and $d = 1.3\,\mu m$.