Optik 166 (2018) 103–109

Contents lists available at ScienceDirect

Optik

journal homepage: www.elsevier.de/ijleo

Original research article

Optical biosensor based on a cladding modulated grating waveguide

Sourabh Sahu[a,*], Jalil Ali[b], Preecha P. Yupapin[c,d], Ghanshyam Singh[a]

[a] Department of Electronics and Communication Engineering, Malaviya National Institute of Technology, Rajasthan, India
[b] Laser Centre, IBNU SINA ISIR, Faculty of Science, Universiti Teknologi Malaysia, 81310 Johor Bahru, Malaysia
[c] Computational Optics Research Group, Advanced Institute of Materials Science, Ton Duc Thang University, District 7, Ho Chi Minh City,700,000, Vietnam
[d] Faculty of Electrical & Electronics Engineering; Ton Duc Thang University, District 7, Ho Chi Minh City, 700,000, Vietnam

ARTICLE INFO

Article history:
Received 19 November 2017
Accepted 5 April 2018

Keywords:
Biosensors
Eigenmode
Optical gratings

ABSTRACT

We have proposed and analyzed a photonic biosensor based on a cladding modulated grating waveguide comprising of the three channels on silicon-on-insulator (SOI) platform. The optical mode is propagated in the center channel and the other two channels interacted with the lateral field. Gratings are designed in the outer channels and parameters are optimized to get a narrow stopband in the transmission spectra. The analysis of the proposed configuration is performed through the eigenmode expansion (EME) method. For a sensing, shift in a resonant band is utilized for a change in a refractive index of a biomaterial. The structure is evaluated for three-waveguide thickness (220 nm, 150 nm, and 100 nm) by considering the fabrication tolerances. The sensor structure with a 100 nm thickness has delivered a higher sensitivity of 322.96 nm/RIU with a limit of detection of the order of 1.03×10^{-4} RIU. A designed sensor is easy to fabricate using existing CMOS fabrication facility and can be used for an application of lab-on-chip devices.

1. Introduction

During the last decade, several types of photonic biosensors have been demonstrated for applications in the field of medical diagnostics, environmental monitoring, chemical detection, biological warfare inspection, etc. Optical sensors provide a fast, reliable, less expensive and easy to use lab on chip (LOC) integration that can outrun the conventional method of medical diagnostics that is laborious and expensive [1]. The most widely used technique for biosensing is based on the fluorescence detection. This technique utilizes the fluorescent labels for prediction of a suitable biomaterial. From this method, the detection up to single molecules can be made, but it suffers from disadvantages such as the variation in the characteristics of the biomaterial due to recombination between the biomaterial and fluorophores, laborious process, and expertise requirement for handling complex equipment [2,3]. A more preferred technique for biosensing is based on the label-free detection, which can bind a suitable biomaterial on the sensor surface through an immobilized layer attached to it. The biorecognition process (binding between analyte and bioreceptor) changes the surface properties of the sensor, which can be characterized to detect the presence of a targeted biomaterial [2]. The bioreceptor is usually a protein (or enzyme) capable of binding a targeted molecule or an analyte.

* Corresponding author.
E-mail addresses: sourabh.ggits@gmail.com (S. Sahu), gsingh.ece@mnit.ac.in (G. Singh).

https://doi.org/10.1016/j.ijleo.2018.04.034
0030-4026/

The inherent feature of the photonic biosensor is its ability to work as a label-free detection system. Most of these optical sensors rely on an interaction between the evanescent field and biomaterial present on the surface of a sensor. Change in the molecular activity causes to alter the refractive index of the biomaterial that simultaneously modifies the effective modal index of a waveguide [4]. This variation of the effective index can be quantified for a label-free detection. Numerous configurations of photonic sensors have been testified based on evanescent field sensing such as interferometer [5–7], ring resonator [8,9], disk resonators [10], photonic crystal waveguide [11,12], surface plasmon resonance [13,14], Bragg gratings [15–17], etc. For instance, the use of interferometer for sensing applications is limited by the sinusoidal dependence of the intensity that causes uncertainty in the detection of biomaterials. The ring and disk resonators based sensors suffer from a lower value of FSR that limits its detection range and additionally having higher bending losses. Apparently higher losses and complexity also restricts the use of photonic crystal for sensing applications whereas the surface plasmon response based devices are highly affected by the deformity in thickness of the deposited material, or the roughness occurred during fabrication [18]. While relating with other resonating devices, the Bragg grating sensor suffers from scattering losses due to the fabrication variability in the corrugation depth of the gratings [18,19]. The configuration provides flexibility for modification of the Q factor of the resonating band by adjusting the geometric parameters.

Bragg waveguide consists of a periodic perturbation of an effective refractive index, which is analogous to a 1D photonic crystal and hence it provides stopband in the transmission spectra. On account of insertion of the phase shift cavity, the Bragg waveguide offers a resonating band at the center of the stopband region [20]. For a biosensing application, the shifting of the resonating band is utilized for quantification of the change in the refractive index of the biomaterial. Juggessur et al. has proposed a device that uses a shifting of the band edge for sensing and demonstrated a sensitivity of the 5.5×10^{-3} nm^{-1} [21]. The phase shift strip Bragg waveguide is reported with a sensitivity of 59 nm/RIU and limit of detection (LOD) of 9.38×10^{-4} RIU [10]. A biosensor based on distributed Bragg reflector by a periodic perforation of the depth of a waveguide is represented in Ref. [20], the sensitivity of 75.8 nm/RIU is reported. As the light confinement is more in the lower refractive index region of the slot waveguide that increases the light-biomaterial interaction. Wang et al. characterized the sidewall gratings at the inner boundaries of the slot waveguide provided an improved sensitivity of 290 nm/RIU [22]. On the contrary, the sidewall gratings are designed at the outer sides of the slot waveguide which shown a sensitivity of 340 nm/RIU [23]. The slot waveguide based resonating structures improved the sensitivity, but it suffers from higher insertion loss. In this work instead of generating a resonating peak in the stopband region, we have demonstrated a structure that delivers a narrow stopband. The proposed configuration consists of three channels, which is designed to lower the interaction between the propagated optical mode at the central core waveguide and the gratings present on the sidewall of the two narrow strip waveguides placed on both sides of it. This configuration has the flexibility to choose a low loss central waveguide, which may have enough length to accommodate the number of gratings that is required for narrowband reflectivity [24–26]. For increasing, the interaction between the evanescent field and biomaterial thinner waveguide (100 nm) is used. The thinner height of waveguide is selected based on the fabrication tolerance; as this dimension is fabricated through a fabrication-step with etch depth of 120 nm as demonstrated by Riccardo et al. [27].

The overall goal is to obtain a biosensing device that is highly sensitive, easy to fabricate using CMOS facility, and can be easily integrated with LOC applications. Here, it is also shown that employing the thinner waveguide increases the presence of optical field increased that result in an increasing value of the sensitivity. The organization of the article is as follows, and Section 2 presents the theoretical modeling of the proposed structure with the alteration of the various design parameters, Section 3 represents the evaluation of the biosensing characteristics followed by conclusion in Section 4 and acknowledgment and references.

2. Modeling of a Cladding Modulated Grating (CMG) waveguide

Fig. 1(a) and (b) schematically represents the proposed structure of the CMG waveguide consists of a central core waveguide having a width of 500 nm and two other waveguides (side channels) of width, SC = 80 nm, separated by gap width (GW). Both the channels have gratings of width ΔW on one side of their sidewalls. The confinement of the field in the CMG waveguide is in the center core region for the fundamental quasi-TE mode. The structure utilizes SOI platform, and it is positioned in between the upper cladding and substrate region both having a thickness of 2 μm using a refractive index of 3.47. Initial waveguide thickness is considered as 220 nm and is drawn using Si material (refractive index 1.444). For an evaluation of the modal effective index full vector based Eigen-mode calculation is performed. The high index contrast between the materials of waveguide and cladding enabled the confinement of the mode in the central core region, and only small fraction of the optical-field (trailing edge of the evanescent field) present outside it, which is interacting with the outer side channels as shown in Fig. 1(c).

This structure is especially compelling because the corrugation in the outer channel provides lower effective index difference between the grating and non-grating region thus provides a weaker coupling strength and hence can provide a narrow stop band [26]. In Fig. 1(a) for a 20 nm corrugation width (ΔW), the effective index difference is around ~0.003, which is very low. The grating period (Λ) is calculated by considering the phase matching condition given as, $\lambda_B = 2 \cdot \Lambda \cdot n_{eff}$, where n_{eff} is the average value of effective refractive index n_1 and n_2. The calculated value of the mode refractive indices are $n_1 = 2.442221$ and $n_2 = 2.439270$ for GW = 100 nm. So the calculated value of Λ is ~317 nm to get a stop band at around 1550 nm wavelength, but for the sake of reducing fabrication variability we consider Λ = 320 nm. The complete device is simulated using eigenmode expansion (EME) method [28], it is a bidirectional method, work well in the frequency domain.