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Subwavelength structures for silicon photonics biosensing

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ABSTRACT

Silicon photonic biosensors hold the potential for highly accurate, yet low cost point-of-care devices. Maximizing the sensitivity of the sensing chips while reducing the complexity and cost of the read-out system is pivotal to realize this potential. Here we present an extensive analysis, both from a practical and a theoretical perspective, of current biosensors, and analyze how subwavelength structures can be exploited to enhance their sensitivity. This study is not restricted just to the near-infrared band as we also determine the sensing capabilities of the suspended silicon waveguides with subwavelength metamaterial cladding working in the mid-infrared range. These waveguides have been recently proposed to cover the full transparency window of silicon ($\lambda <\sim 8.5$ μm), where the fingerprint spectral region of many molecules takes place and so a plethora of evanescent field absorption-based applications will be developed in the near future.

1. Introduction

Silicon photonic biosensors are capable of detecting trace amounts of biomolecules, such as antibodies and proteins, and can monitor their reactions in real-time, without prior labeling of the targets [1,2]. Combined with their ability to detect several different analytes in parallel in a single chip, this makes them ideally suited for early diagnosis of diseases [3–5] and drug discovery [6,7]. They are also considered promising candidates for the development of lab-on-chip platforms and point-of-care devices. The basic principle underlying their operation is evanescent field sensing, illustrated in Fig. 1. The surface of an optical waveguide is functionalized with receptor biomolecules, which bind, with high specificity, to the analyte [8], e.g. antibodies that will attach only to their corresponding antigens. When an aqueous solution containing the analyte flows over the waveguide, the analyte will bind to the receptors on the waveguide surface, locally changing its optical properties. Light propagating through the waveguide is confined by total internal reflection, but 'senses' the medium surrounding the waveguide through the evanescent tails of its electric field. When a light-wave interacts with the biomolecules that are binding to the waveguide surface some of its properties (wavelength, amplitude or polarization) change, and by monitoring these changes the analyte can be detected. Thus, while the specificity of photonic biosensors depends mainly on the surface functionalization, their sensitivity strongly depends on their optical implementation. Indeed, over the last decade, extensive research efforts have been devoted to optimizing this sensitivity: different waveguide types, such as silicon wire waveguides, slot waveguides and more recently subwavelength grating (SWG) waveguides have been explored, as well as different sensing architectures, mainly based on ring-resonators and Mach-Zehnder interferometers have been studied.

In this paper we aim to provide both theoretical and practical insight into photonic sensor design, with a particular emphasis on subwavelength structures which can provide some of the highest sensitivities to date. To this end, we start with a systematic description of the waveguide and architectural parameters that govern sensitivity in Section 2. In Section 3, we describe the practical realization of two complete sensing systems: one based on ring-resonators, and one based on interferometry. In Section 4, we systematically analyze, for the first time, the sensitivity of subwavelength grating waveguides, revealing that current design parameters may be sub-optimal. The sensing properties of suspended silicon waveguides, operating in the mid-infrared wavelength range, are discussed, for the first time, in Section 5. Finally we present some concluding remarks.

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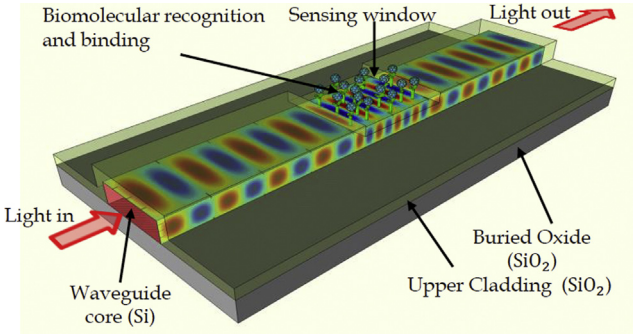

Fig. 1. Photonic wire waveguide for biosensing in the silicon-on-insulator platform.

2. Fundamentals of photonics biosensors

This section lays the theoretical foundation required to understand the parameters that govern the sensitivity and limit of detection of photonic biosensors.

2.1. Waveguide and architecture sensitivity

Fig. 1 illustrates a typical biosensor waveguide in silicon-on-insulator. The waveguide is covered by a protective SiO_2 cladding, except for the sensing window, where the waveguide core is exposed to the surrounding medium. When a sample of the analyte is delivered to the sensing window through a microfluidic channel (not shown in the figure) and gets in touch with the functionalized surface of the waveguide, molecular binding takes place. Interaction of these molecules with the evanescent tail of the guided mode field changes its effective index and thus its wavelength [9]. One of the most important characteristics of a sensor is the waveguide mode sensitivity S_w which maps the physical change due to molecular binding, into effective index variations:

$$S_w = \frac{\partial n_{\text{eff}}}{\partial \Gamma}, \tag{1}$$

where $\partial\Gamma$ is the variation of any physical parameter. Unfortunately, the effective index of a waveguide mode is not a directly measurable quantity, so in order to be useful, they must be mapped into a quantity that can be readily detected. This is achieved by using a photonic sensing architecture. These architectures can be broadly categorized in two different configurations: interferometric and resonant [10]. In interferometric architectures the effective index variations are mapped into an optical phase shift, $\Delta\varphi$, while in a resonant architectures they are mapped into a wavelength shift $\Delta\lambda$. Focusing on the interferometric type of sensor, the architecture sensitivity S_a can be defined as

$$S_a = \frac{\partial \varphi}{\partial n_{\text{eff}}}, \tag{2}$$

and the total photonic device sensitivity S can be calculated as the product of waveguide and architecture sensitivities

$$S = S_a S_w = \frac{\partial \varphi}{\partial n_{\text{eff}}} \frac{\partial n_{\text{eff}}}{\partial \Gamma} = \frac{\partial \varphi}{\partial \Gamma}. \tag{3}$$

While this magnitude depends only on the photonic integrated circuit, the limit of detection (LOD), i.e. the minimum amount of detectable variation in the physical parameter $\Delta\Gamma_{min}$, will also depend on the minimum detectable phase shift $\Delta\varphi_{min}$ that can be accurately resolved by the measurement apparatus. This quantity is sometimes referred as the set-up resolution R and can be related to the system noise variance σ_φ through [11]:

$$R = \Delta\varphi_{min} = 3\sigma_\varphi. \tag{4}$$

From these definitions the LOD can be easily calculated as

$$LOD = \Delta\Gamma_{min} = \frac{R}{S_a S_w}. \tag{5}$$

The same type of definition applies to resonant sensors by substituting, in Eqs. (2)–(4), φ by λ.

Since the LOD depends both on the photonic chip and on the resolution of the measurement apparatus, it is difficult to compare the performance of different sensor devices using this metric. Researchers working on resonant sensors therefore make a distinction between the system LOD (sLOD) which depends on the complete set up, and the intrinsic LOD (iLOD) which only depends on the photonic device itself [12]. This distinction will be explained in the Section 2.3. Unfortunately, to the authors' knowledge, no such metrics have been proposed for interferometric biosensors.

2.2. Bulk and surface waveguide sensitivities

Two different waveguide sensitivities are defined in the literature: bulk $S_{w,\,bulk}$ and surface sensitivity $S_{w,\,surf}$.

Referring to Fig. 2(a), bulk sensitivity is defined as the ratio of change of the mode effective index (∂n_{eff}) and the change of the refractive index of the material covering the waveguide (∂n_c)

$$S_{w,\,bulk} = \frac{\partial n_{\text{eff}}}{\partial n_c}\ (\text{RIU/RIU}). \tag{6}$$

Surface sensitivity is defined as the ratio of the mode effective index change (∂n_{eff}) and the change in thickness (∂t) of the adsorbed molecular layer, as shown in Fig. 2(b)

$$S_{w,\,surf} = \frac{\partial n_{\text{eff}}}{\partial t}\ (\text{RIU/nm}). \tag{7}$$

These are purely electromagnetic definitions which are very useful for photonic designers. From the chemical point of view, two related magnitudes can be used. Waveguide bulk sensitivity can be also defined as the ratio of effective index variation and the change in the analyte concentration (∂c, in moles per liter or M):

$$S_{w,\,bulk} = \frac{\partial n_{\text{eff}}}{\partial c}\ (\text{RIU/M}). \tag{8}$$

This measure is only of relative importance for the biomolecular recognition capability of the sensor (which takes place in the sensor surface) but is sometimes used as an intermediate step in sensor characterization. On the other hand waveguide surface sensitivity can be also defined as

$$S_{w,\,surf} = \frac{\partial n_{\text{eff}}}{\partial \rho_s}\ \left(\frac{\text{RIU}}{\text{pg/mm}^2}\right), \tag{9}$$

where ρ_s is the mass surface density of the adsorbed layer.

Please notice that in all these definitions, it is always assumed that there is enough analyte to completely fill up (cover) all the volume (surface) of the sensing window. For other applications, in which there is a very limited amount of analyte (for example, single molecule detection), other metrics should be used.

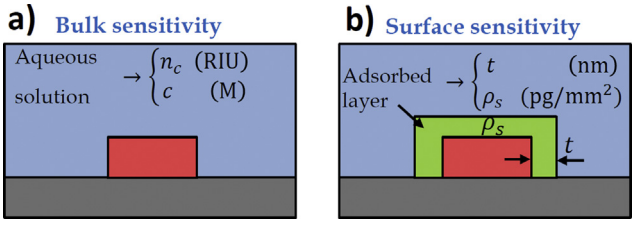

Fig. 2. Bulk and surface sensitivities.