Surface Desensitization of Polarimetric Waveguide Interferometers

Colin Worth, Bennett B. Goldberg, *Member, IEEE*, Michael Ruane, *Senior Member, IEEE*, and M. Selim Ünlü, *Senior Member, IEEE*

***Abstract*—Nonspecific surface binding of small protein molecules presents a major obstacle to surface biosensing techniques attempting to detect very low concentrations (< 1 pg/mm^2) of large biological objects such as cells and bacteria. A new method for selective desensitization of a polarimetric waveguide interferometer eliminates the background noise from nonspecific surface binding. We demonstrate the ability to tune the phase sensitivity of a waveguide interferometer as a function of the distance of the biological or chemical analyte from the waveguide surface. This makes possible a sensor that has zero sensitivity at a particular distance where nonspecific surface binding occurs without significantly reducing the sensitivity to target larger biological species.**

***Index Terms*—Evanescent wave biosensing, surface sensing, waveguide interferometry.**

I. Introduction

OPTICAL techniques for detecting low concentrations of cells or bacteria in solution have become popular over the past few years [1], [2]. In particular, waveguide interferometry has the potential to become an effective approach for biosensing [3], [4], [10]–[13]. Waveguide interferometric biosensing exploits the differential change in the phase velocity of the guided modes of a waveguide caused by biological particles bound to the surface. While other optical biosensing techniques, e.g., total internal reflection fluorescence [5], [6], tag particles and sense the resulting fluorescence, waveguide interferometry detects any particle bound to the surface though a change in the index of refraction. Thus, the selectivity of the surface binding chemistry determines how well the biosensor can distinguish small numbers of the desired particle type. Perfecting selective chemical binding techniques for specific agents is a major challenge facing biosensors.

This paper describes a novel purely optical method of distinguishing between specifically and nonspecifically bound particles based on particle size and distance above the waveguide surface. In most biosensing applications, specifically bound particles—such as bacteria—are much larger (several micrometers across) than nonspecifically bound particles—typically proteins (several nanometers across). We show that tuning the evanescent field of the polarization modes desensitizes a thin layer (20–30 nm) above the waveguide surface, reducing the response to nonspecific binding by a factor of one hundred or more.

Manuscript received June 6, 2001.

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Publisher Item Identifier S 1077-260X(01)11236-0.

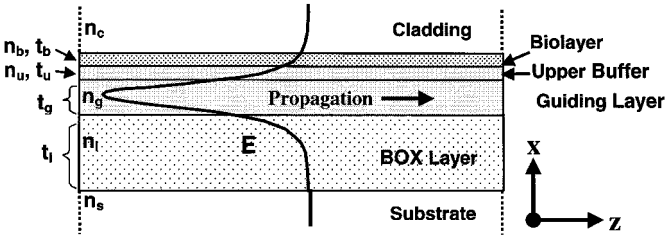

Fig. 1. Generic slab waveguide structure. Thicknesses t_x and refractive indexes n_x in each layer vary. BOX is buffered oxide layer and forms first buffer layer. Typical thickness values: $t_l \simeq 1500$ nm, $t_g \simeq 150$ nm, $t_u \simeq 5$ nm, $t_b \simeq 10$ nm. TE_0 mode electric field profile displayed. Mode propagates in $+z$ direction.

II. Mode Behavior of Evanescent Waveguide Interferometers

A. Waveguide Modes

A slab waveguide consists of multiple dielectric layers, each with different refractive index, stacked along the x dimension (Fig. 1) to form a guiding layer bounded by upper and lower buffer layers. The slab has infinite extent in the $\vec{y}$ and $\vec{z}$ dimensions. The guiding layer supports a propagating light wave a stationary sinusoidal amplitude profile in the x direction and exponentially decaying evanescent fields into the buffer layers (Fig. 1). The electric field in the guiding layer is given by

$$E = \cos(k_x x) e^{ik_z z - i\omega t}. \tag{1}$$

The total wave vector of the light $\vec{k}$ is determined by the index of refraction of the guiding medium according to

$$k_x^2 + k_z^2 = n_g^2 k_0^2 \tag{2}$$

where $k_0 = 2\pi/\lambda_0$ is the free-space wave vector. The boundary conditions at each interface require the z component of the wave vector to be conserved. k_z is called the propagation constant β of the mode. A fundamental property of waveguides is that mode solutions exist only for particular discrete values of β. The guiding layer thickness t_g is chosen to allow only two orthogonally polarized propagating modes TE_0 and TM_0 (electric and magnetic field, respectively, parallel to the waveguide dielectric layer interfaces). The guiding layer thickness t_g is typically on the order of 100 nm for $\lambda = 633$ nm.

B. Measuring Changes in the Propagation Constant of Waveguide Modes

Interferometry measures relative phase shifts between two coherent beams of light by observing their interference fringe pattern. Any change in the optical path length of one of the beams causes a resultant shift in the interference pattern. In polarimetric waveguide interferometry, the two beams correspond to the two lowest order waveguide modes TE_0 and TM_0. Binding of biological particles to the waveguide surface changes the index of refraction of a thin layer above the surface, leading to a change in the mode propagation constants. The change in the propagation constants leads to a change in the optical path length of the light and a total phase shift $\Delta\phi$ proportional to the average change in the propagation constant per unit length $\Delta\beta$ times the interaction length L.

$$\Delta\phi = \Delta\beta L. \tag{3}$$

When the two modes are interfered, the overall phase shift in the interference pattern is equal to the *relative* phase shift between the two modes

$$\Delta\phi = \Delta\phi_{\mathrm{TM}} - \Delta\phi_{\mathrm{TE}} = (\Delta\beta_{\mathrm{TM}} - \Delta\beta_{\mathrm{TE}})L. \tag{4}$$

C. Formulation of Sensitivity

Modal phase shifts due to binding of biological particles to the waveguide surface can be estimated by representing the biological matter by a uniform dielectric layer located above the waveguide. For widespread binding of many small particles, as in nonspecific binding of proteins, this uniform model is appropriate.

Interferometric sensitivity is defined as the change in the modal propagation constant β per unit change in biolayer index n_b [3], [10]. The response to changes in the biolayer index is determined by numerically solving for the change in propagation constants [8], β_{TM} and β_{TE} as a function of the change Δn_b. The sensitivity of each mode is

$$S_{\mathrm{TE}} = \frac{\Delta n_{\mathrm{eff;TE}}}{\Delta n_b} \tag{5}$$

$$S_{\mathrm{TM}} = \frac{\Delta n_{\mathrm{eff;TM}}}{\Delta n_b} \tag{6}$$

where n_{eff} is the effective index or normalized propagation constant

$$n_{\mathrm{eff}} = \frac{\beta}{k_0}. \tag{7}$$

The differential sensitivity ΔS is then

$$\Delta S = S_{\mathrm{TM}} - S_{\mathrm{TE}}. \tag{8}$$

III. Results and Discussion

Both the TM_0 and TE_0 modes respond to refractive index changes near the waveguide surface. However, the interferometric measurement signal is the *relative* change between the two phase shifts. To maximize interferometric sensitivity, we vary the guiding layer thickness to maximize the difference between the responses of each mode to a change in biolayer index. Fig. 2 shows the sensitivity of the TE_0 and TM_0 modes as a function of t_g, the thickness of the guiding layer for a materials system consisting of a silicon nitride–silicon dioxide waveguide ($n_l = n_u = 1.465$, $n_g = 2.02$) operating at $\lambda = 632.8$ nm and a biolayer thickness t_b of 10 nm at the waveguide surface. The biolayer index is varied around $n_b = 1.333$, the index of water, to determine the sensitivities. The maximum differential sensitivity occurs at $t_g \simeq 166$ nm. A similar value of t_g gives the maximum sensitivity for thicker biolayers as well due to the fact that the TE_0 mode is significantly more confined within the guiding layer than the TM_0 mode.

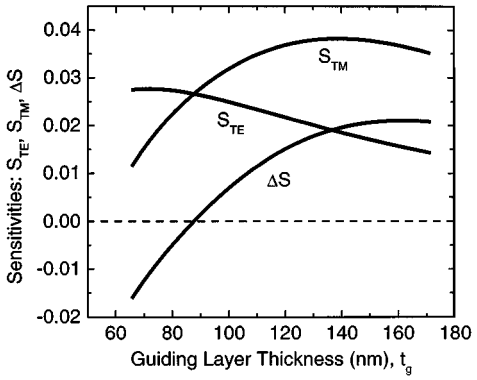

Fig. 2. Waveguide sensitivities versus guiding layer thickness for a silicon nitride–silicon dioxide waveguide ($n_l = n_u = 1.465$ nm, $n_g = 2.02$ nm, $t_l = 1500$ nm, $t_u = 4$ nm), operating at $\lambda = 632.8$ nm and a biolayer thickness $t_b = 10$ nm.

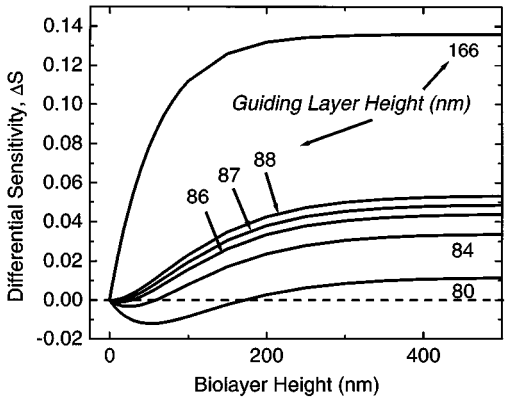

Fig. 3. Waveguide sensitivity versus biolayer height for several guiding layer thicknesses. $\Delta n_b = 0.005$.

Fig. 3 shows the variation of the differential sensitivity ΔS with the thickness of the biolayer t_b for several values of t_g, including $t_g = 166$ nm. The sensitivity saturates with increasing biolayer thickness at around half a wavelength (300–400 nm) because the evanescent field strength has decayed at this distance. For lower values of t_g, Fig. 3 shows an unexpected result. Not only do the steepness and the saturation point of the sensitivity curves change, but the curvature changes as well. For a guiding layer thickness of $t_g = 84$ nm the differential sensitivity is *negative* (the TE mode is more sensitive than the TM mode) for a biolayer thickness t_b up to 60 nm. At a biolayer thickness of $t_b = 60$ nm, the TE and TM modes respond equally

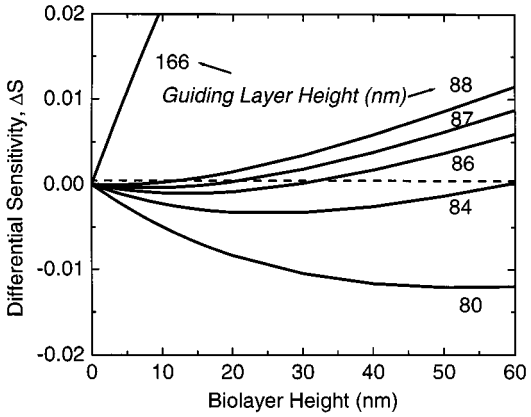

Fig. 4. Waveguide sensitivity versus biolayer thickness, enlarged for thin biolayers for several guiding layer thicknesses. For $t_g = 87$ nm, $|\Delta S| < 10^{-4}$ for $0 < t_b < 25$ nm. Compare to ΔS for $t_g = 166$ nm.

to changes in the biolayer index. Thus, although the response of each mode is finite, the differential response is zero. This represents the new desensitization effect for thin biolayers that we have discovered.

Fig. 4 shows an enlarged plot of Fig. 3 as a function of biolayers thickness. Although the saturated differential sensitivity (important for detection of large particles) is reduced by a factor of 60%, Fig. 4 shows that the biosensor is essentially blind to binding of small particles from a few ångströms to 20 nm in thickness. A comparison of the curves for $t_g = 87$ nm and $t_g = 166$ nm shows that the average sensitivity to small particles in the range 5 nm $< t_b <$ 20 nm has been reduced by a factor of 100 and proteins or cells as large as 30 nm produce a signal almost 20 times less than they would have in a nondesensitized system.

The sensitivity of a waveguide mode to a particular uniform biolayer is related to the proportion of the mode intensity contained in the evanescent fields interacting with the biolayer. Fig. 5(a) compares the normalized TM_0 mode profiles for a waveguide optimized for high differential sensitivity ($t_g = 166$ nm) to a waveguide optimized for desensitization ($t_g = 87$ nm). Fig. 5(b) compares the TE_0 mode profiles. The major difference between the TE_0 and the TM_0 modes is the large electric-field discontinuity for the TM_0 mode at the guiding layer boundary. As the thickness of the guiding layer shrinks, the TE_0 mode barely changes, while the TM_0 mode becomes significantly less confined as shown in Fig. 5(a). The weakening confinement increases the relative mode weight in the evanescent field, but decreases the mode weight in the neighborhood of the waveguide surface. At a guiding layer thickness of $t_g = 87$ nm, the relative field intensities of TE_0 and TM_0 to their total mode intensities are identical at 21 nm above the waveguide. Thus, the change in the optical path length is identical, leading to zero sensitivity at this point.

IV. Conclusion

Polarimetric waveguide interferometers use the differential phase shift between two orthogonal waveguide modes to detect biomaterials bound to the waveguide surface. The differential

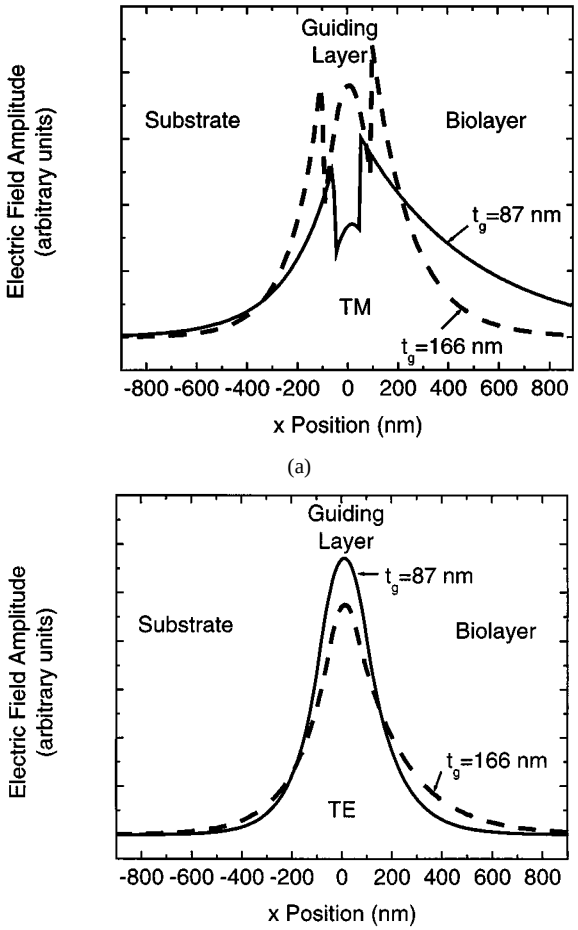

Fig. 5. TE_0 and TM_0 mode electric field profiles for guiding layer thicknesses of 166 and 87 nm. (a) TM_0 mode. (b) TE_0 mode. $x = 0$ in center of guiding layer. Asymmetry is due to unequal buffer thicknesses.

interference technique and the complex response of waveguide modes to changes in waveguide thickness allow us to tailor the optical phase response of the biosensor to discriminate against nonspecific binding of small molecules such as proteins. Interferometric optical sensors are fundamentally limited not by their optical phase resolution, but by chemical processes at the waveguide/biolayer interface, especially nonspecific binding of proteins. We believe that the method of surface desensitization reported here will allow results several orders of magnitude better than previously reported for the detection of large molecules such as bacteria bound to a waveguide surface. We are also implementing a doubly differential technique to compensate for thermal and concentration gradients within the waveguide and sample. The latter effects may play a greater role once the signal due to nonspecific binding is reduced by desensitization.

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