Near ultraviolet-wavelength photonic-crystal biosensor with enhanced surface-to-bulk sensitivity ratio

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We report the design and characterization of biosensors based on one-dimensional photonic-crystal reflectance filters operating at near-ultraviolet wavelengths. Rigorous coupled-wave analysis simulations predict an increased confinement of the resonant electric field at the surface of this biosensor as compared to previously fabricated near-infrared photonic-crystal biosensors. This change in the resonant electric field provides an improvement of over 4.5 times in the surface-sensitivity to bulk-sensitivity ratio, and therefore enables enhanced detection resolution for biomolecules adsorbed on the biosensor surface. These biosensors will be especially important for applications requiring the detection of small molecules or ultralow analyte concentrations. © *2006 American Institute of Physics*. [DOI: 10.1063/1.2219984]

There is a growing interest in biomolecular detection platforms for drug discovery,[1] environmental detection, medical diagnostics, and life science research.[2] Traditional labeled detection methods can introduce significant experimental complexity, uncertainty, and cost. Label-free biosensors are an important emerging class of sensors that circumvent these limitations and can reveal binding affinity, specificity, and kinetics.[3]

A near-infrared (n-IR) photonic-crystal (PC) optical biosensor has previously been demonstrated[4] for label-free detection of biochemical interactions, exhibiting a mass density sensitivity resolution of less than 1 pg/mm^2, refractive index discrimination down to 10^{-6} refractive index units (RIU), and a large dynamic range.[5] Recently, we have discovered that by substituting a low refractive index (RI) porous dielectric for the UV-cured polymer used in previous designs, the sensitivity of the PC biosensor is significantly enhanced.[6] In the present work we demonstrate the scaling of this process to fabricate near-ultraviolet (n-UV) PC biosensors.

The label-free PC optical biosensor uses an optically transparent substrate and a low refractive index nanostructure onto which a high refractive index material is deposited, as shown in Fig. 1(a). The device is designed as a reflectance filter with unity reflection efficiency at band center, the precise reflection characteristics of which are governed by the height and period of the surface structure, the thickness of the high index (TiO_2) layer, and the strength of modulation (RI difference of the materials). The reflection arises by utilizing a guided-mode resonance effect,[7] in which a structured surface incorporates a subwavelength periodic dielectric permittivity modulation.[8] In the structures reported here, the refractive index modulation is designed to produce a photonic band gap in the direction parallel to the substrate surface, so as to prevent lateral propagation of light coupled to the structure at the resonant wavelength. The subwavelength periodic structure supports a zeroth-order mode in the direction normal to the substrate surface that results in reflection efficiency near 100% at the resonant wavelength, while nonresonant wavelengths may transmit through the structure. The reflection maximum for the n-IR and n-UV biosensors is centered at 795 and 405 nm, respectively. The design and fabrication of plastic n-UV PC reflectance filters has been described in our prior work[9] and the process therein is extended to a low RI porous dielectric to fabricate these n-UV biosensors. In review, a one-dimensional surface structure (period=250 nm) is patterned in a Si/SiO_2/poly (methylmethacrylate) substrate by electron-beam lithography and etched into the SiO_2 using reactive ion etching. This structure is used as a "master" mold from which a stamp in polydimethylsiloxane (PDMS) is prepared. This PDMS stamp is used in a nanoreplication process to pattern a glass substrate coated with a low RI spin-on glass (Nanoglass, Honeywell). This structure is shown in Fig. 1(b). Finally, a layer of high RI TiO_2 is evaporated onto the patterned surface to obtain the biosensor device.

The spectral location of peak reflection, or peak-wavelength value (PWV), is readily tuned by changes in the optical density of the medium atop the sensor surface lying within the range of the evanescent electric field. Therefore, bulk RI changes of the cover medium will induce a shift in the PWV, as will any thickness or density changes of a surface-bound biomolecular layer. Since bulk solution RI variations are a significant source of noise for surface-based optical biosensors, a higher ratio of surface-to-bulk sensitivity will consequently yield enhanced detection resolution. We therefore seek to maximize the PWV shift in response to a given biomolecular layer while simultaneously limiting sensitivity to RI variations of the bulk media.

By modifying the device structure and, hence, the evanescent electric field profile, one can expect to change the sensitivity characteristics of the biosensor. By scaling down the n-IR biosensor and thereby enabling operation at lower wavelengths, we expect to reduce the extent of the evanescent electric field since the field extent is proportional to the resonant wavelength for a given angle of incidence and set of material properties. Computer simulations using the rigorous coupled-wave analysis[10] (RCWA) method for an optimized device geometry give the spatial profile of the electric field in the x-z plane for the n-UV and n-IR biosensors in Figs.

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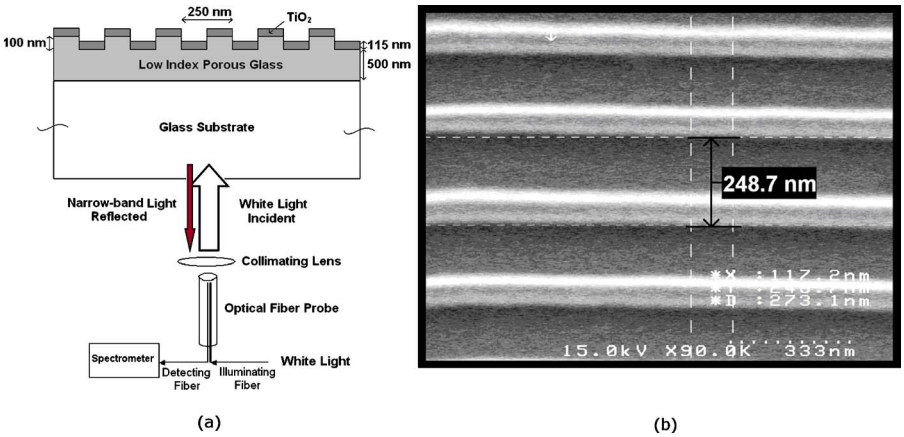

FIG. 1. (Color online) (a) Cross-section schematic of the n-UV PC biosensor and measurement setup and (b) scanning electron microscopy image of the one-dimensional periodic surface structure as replicated into low RI porous spin-on glass.

2(a) and 2(b), respectively. The plot shows the electric field intensity profile for one period of the surface structure, where periodic boundary conditions are applied to the left and right limits and a grid size of 1 nm × 1 nm is used. The device is illuminated with a TM polarized (i.e., the light is polarized perpendicular to the y axis, the length of the surface structure) plane wave, incident from below the device. With respect to the scale shown in the figure, this plane wave has unit amplitude. It is readily apparent that the field is more confined to the biosensor surface in the n-UV biosensor [Fig. 2(a)], and we can therefore expect changes in bulk refractive index to less dramatically tune the PWV.

To test the performance of the biosensor, the bulk sensitivity was determined by measuring the change in PWV as the RI of the material covering the sensor varied. Figure 3(a) shows the change of PWV for air, water and isopropyl alcohol (IPA) applied to the sensor surface by placing a drop of each under a glass cover slip. The bulk-shift coefficient (defined as $\Delta\mathrm{PWV}/\Delta n$ where n is the refractive index of the bulk medium) is given by the slope of this line and quantifies the bulk sensitivity of the biosensor.

Sensitivity to surface-adsorbed material was characterized by the detection of a single layer film of Poly(Lys, Phe) (PPL) (Sigma-Aldrich; MW=35 400 Da) prepared to a 0.5 mg/ml solution with 0.01 M phosphate buffered saline (PBS) (pH=7.4) applied to the sensor surface. The bioadhesion test commenced with the pipetting of PBS onto the sensor surface to establish a baseline PWV. After 10 min, the buffer was replaced with PPL solution and was allowed to stabilize for 30 min. The kinetics for this process is shown in Fig. 3(b).

The highly confined electric field at the surface of the n-UV biosensors, as predicted by RCWA computer simulations, results in a very low bulk shift coefficient of 49.1 nm/RIU. In contrast, the previously fabricated n-IR biosensors had a bulk shift coefficient of 302 nm/RIU.[6] The PWV change in response to a PPL monolayer deposited on the n-UV biosensor was measured to be 3.19 nm, whereas for the n-IR biosensors it has been previously measured to be 4.07 nm.[6] To provide a quantitative comparison of the surface-to-bulk sensitivity for the two biosensors, we divide the surface shift by the bulk shift coefficient, which results in the values 0.065 and 0.013 for the n-UV and n-IR biosensors, respectively. This translates to an improvement of over 4.5 times for the surface-to-bulk sensitivity in the n-UV biosensor over the n-IR biosensor.

The lowered extent of the electric field at the surface of biosensors operating in a low wavelength regime serves to considerably increase the surface-to-bulk sensitivity ratio. This figure of merit correlates directly with the PC biosensor's limit of detection. Therefore, the n-UV biosensor presented in this work should enable improved biomolecular detection at low concentrations and is a step forward towards the ultimate goal of single molecule resolution.

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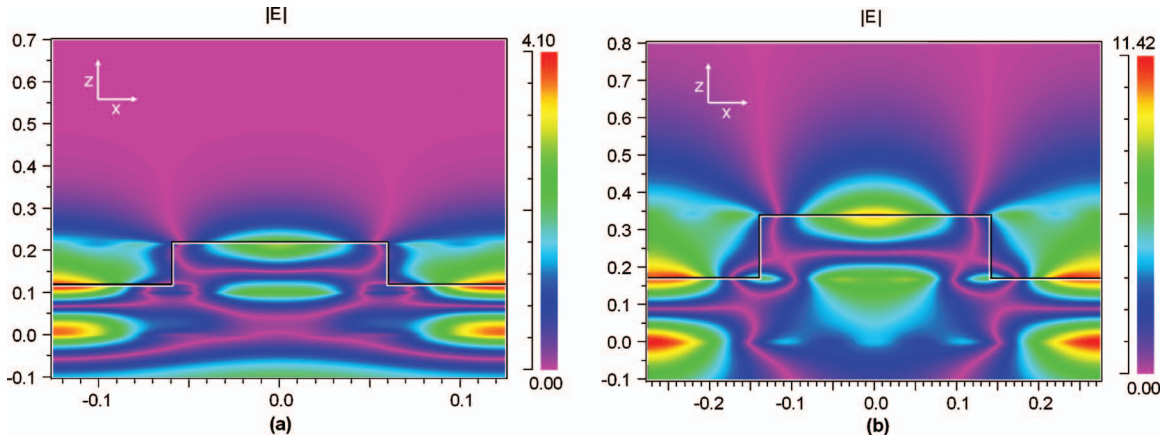

FIG. 2. (Color) RCWA simulations of resonant electric field profile for the (a) n-UV biosensor (λ=405 nm) and (b) n-IR biosensor (λ=795 nm). The field is seen to be more tightly confined to the surface of the n-UV biosensor. The plane-wave light source excites the structure from below along the z axis and has unit amplitude. The x and z axes indicate the extent of the simulation region, in microns.

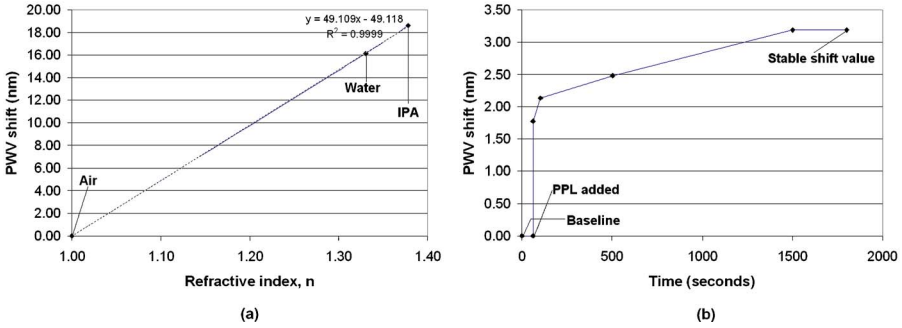

FIG. 3. (Color online) Experimental results of the (a) bulk and (b) surface sensitivity experiments. The bulk shift coefficient is calculated by the slope of the line (49.1 nm/RIU) that is obtained by plotting the PWV shift as the RI of the material at the surface of the biosensor is changed from air (RI=1) to that of water (RI=1.33) and IPA (RI=1.38), and the maximum surface shift for a layer of PPL deposited at the biosensor surface is 3.19 nm.

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[1]D. G. Myszka and R. L. Rich, Pharm. Sci. Technol. Today **3**, 310 (2000).
[2]M. Malmqvist and R. Karlsson, Curr. Opin. Chem. Biol. **1**, 378 (1997).
[3]A. J. Cunningham, *Introduction to Bioanalytical Sensors* (Wiley, New York, 1998), p. 418.
[4]B. T. Cunningham, J. Qiu, P. Li, J. Pepper, and B. Hugh, Sens. Actuators B **85**, 219 (2002).
[5]B. T. Cunningham, P. Li, S. Schulz, B. Lin, C. Baird, J. Gerstenmaier, C. Genick, F. Wang, E. Fine, and L. Laing, J. Biomol. Screening **9**, 481 (2004).
[6]I. D. Block, L. L. Chan, and B. T. Cunningham, Sens. Actuators B (to be published).
[7]S. S. Wang, R. Magnusson, and J. S. Bagby, J. Opt. Soc. Am. A **7**, 1470 (1990).
[8]S. S. Wang and R. Magnusson, Appl. Opt. **32**, 2606 (1993).
[9]N. Ganesh and B. T. Cunningham, Appl. Phys. Lett. **88**, 071110 (2006).
[10]M. G. Moharam and T. K. Gaylord, J. Opt. Soc. Am. **71**, 811 (1981).