Research Article

Sensitive Label-Free Biomolecular Detection Using Thin Silicon Waveguides

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We review our work developing optical waveguide-based evanescent field sensors for the label-free, specific detection of biological molecules. Using high-index-contrast silicon photonic wire waveguides of submicrometer dimension, we demonstrate ultracompact and highly sensitive molecular sensors compatible with commercial spotting apparatus and microfluidic-based analyte delivery systems. We show that silicon photonic wire waveguides support optical modes with strong evanescent field at the waveguide surface, leading to strong interaction with surface bound molecules for sensitive response. Furthermore, we present new sensor geometries benefiting from the very small bend radii achievable with these high-index-contrast waveguides to extend the sensing path length, while maintaining compact size. We experimentally demonstrate the sensor performance by monitoring the adsorption of protein molecules on the waveguide surface and by tracking small refractive index changes of bulk solutions.

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1. INTRODUCTION

Sensing techniques employing fluorescent labeling of molecules are commonly used for simple binding event detection to determine the presence or absence of molecular targets in solution. However, there are few tools available that are capable of providing quantitative, label-free, and real-time monitoring of molecular interactions. There has been extensive activity in both the academic and commercial sectors to develop new sensing technologies that offer these capabilities, allowing important information such as molecular concentrations, binding affinities, and rate constants to be obtained. There have been numerous proposed solutions using planar waveguide optics such as surface plasmon resonance (SPR) [1], ridge waveguide evanescent field sensors [2–6], waveguide-based microcantilevers [7, 8], porous silicon resonant structures [9], optofluidic waveguides [10] as well as many others. The two methods that have been the most widely studied are the SPR sensors and the ridge waveguide evanescent field sensors. These methods employ a waveguide structure as the transducer and surface functionalization chemistry to immobilize receptor molecules on

the sensor surface. The receptors capture target molecules thereby accumulating material on the waveguide surface, which serves to induce a phase shift of the propagating optical field. In the case of SPR sensors, the optical field is the surface plasmon mode which propagates along the metal-analyte interface, whereas for the planar waveguide evanescent field sensor it is the evanescent tail of the optical waveguide mode. The induced phase shift of these devices can be monitored in real time to determine the density of molecules attached to the waveguide surface.

Although only SPR sensors have become widely commercially available, there is significant interest in developing planar waveguide evanescent field sensors due to the potential advantages offered by this technology. For example, these sensors are expected to provide increased sensitivity over SPR due to the much lower optical transmission losses [1]. It is possible for light to propagate through planar optical waveguides for distances of several centimeters with little loss, whereas for the surface plasmon devices the field is effectively extinguished after propagating a few tens of micrometers. This allows a much larger interaction length of the light with the biomolecules to be achieved, thereby

FIGURE 1: Illustration of a waveguide evanescent field sensor. Molecular targets are captured by receptor molecules at the waveguide surface thereby changing the waveguide mode effective index. This creates a phase shift of the propagating optical mode.

allowing a larger phase shift to be observed in the sensor output. In addition, the planar waveguide evanescent field sensor is well suited for arraying permitting the development of high density, waveguide-based microarray biochips for multianalyte monitoring.

Ridge waveguide evanescent field sensors have been demonstrated in many planar material systems such as silica [2], polymer [3], and silicon nitride [4, 5] as well as nonplanar systems including optical fibers [11]. The planar devices offer many well-known advantages over the nonplanar approaches including the possibility of creating arrays of sensors on a compact chip, the ability to integrate additional optical and electrical components on chip thereby increasing functionality, the capability for mass production using high volume semiconductor manufacturing techniques, and the possibility to integrate microfluidic systems on chip permitting convenient analyte delivery and reduced consumed sample volumes. Waveguide evanescent field sensors have been demonstrated using various waveguide circuit configurations including grating couplers [12], Mach-Zehnder interferometers [2, 3, 5], ring resonators [4, 6], and microdisks [13, 14]. It has been shown that the sensitivity achievable with these devices improves with increasing refractive index contrast between the waveguide core and the surrounding cladding materials [15, 16]. Motivated by these predictions, as well as other important advantages that will be discussed, we have developed evanescent field sensors using the silicon-on-insulator (SOI) material platform. The SOI material system consists of a top silicon waveguide core layer positioned on a silicon dioxide lower cladding layer grown on a silicon substrate [17] and has the highest refractive index contrast of all commonly available planar waveguide systems.

In this manuscript, we review our work developing evanescent field sensors in SOI. We show that silicon nanophotonic waveguides with dimensions smaller than the wavelength of light they guide have a strong evanescent field at the waveguide surface for the transverse magnetic (TM) polarized waveguide mode. This results in increased interaction with surface bound molecules, which leads to larger response compared to waveguide sensors fabricated on

traditional low-index-contrast material platforms [16, 18]. Furthermore, we show that using submicrometer dimension silicon waveguides with rectangular cross section, known as silicon photonic wires, allows extremely compact devices to be realized using well-established silicon fabrication technology. The compact footprint of these sensors allows two-dimensional sensor arrays to be fabricated allowing both spotter-based functionalization and microfluidic delivery systems to be used. We present experimental results demonstrating the performance of these devices using two different waveguide configurations: the Mach-Zehnder interferometer (MZI) and the ring resonator. We illustrate the performance of these sensors to track homogeneous refractive index change of bulk solutions and to monitor the formation of thin molecular layers on the waveguide surface.

2. THEORY: HIGH-INDEX-CONTRAST WAVEGUIDES FOR EVANESCENT FIELD SENSING

A schematic of a planar waveguide evanescent field sensor is shown in Figure 1, where specific target molecules are captured by receptor molecules on the waveguide surface. This induces an effective index change of the propagating optical mode, causing a phase shift to be observed at the device output. A primary goal in the design of these devices is to maximize their sensitivity for the application of interest. These applications can be divided into two categories including homogeneous sensing, which involves a uniformly distributed analyte extending over a distance well exceeding the evanescent field penetration depth, and surface sensing in which the analyte is bound to the waveguide surface in a layer that is usually much thinner than the evanescent field penetration. The intrinsic sensitivity of the devices is determined by the overlap of the squared waveguide mode field with the analyte. For homogeneous sensing, the intrinsic sensitivity is defined as the change of effective index ∂N_{eff} of the waveguide mode relative to homogeneous refractive index change of the bulk solution ∂n_c : $\partial N_{\text{eff}}/\partial n_c$. For the case of surface sensing, it is defined as the effective index change relative to the variation of thickness d of an adsorbed molecular layer with a given refractive index: $\partial N_{\text{eff}}/\partial d$ [12]. These sensitivity constants are independent of waveguide length and are used to compare waveguides of different cross-sectional geometries and refractive index profiles.

From the variational theorem of waveguides [19], ∂N_{eff} caused by a local change of dielectric constant $\Delta\epsilon(x, y)$ can be expressed as

$$\partial N_{\text{eff}} = c \int \Delta\epsilon E \cdot E^* dx dy, \quad (1)$$

where $E(x, y)$ is the normalized modal electric field vector. The sensitivity of the device thus scales with the squared amplitude of the electric field within the perturbation, and therefore with the fraction of the modal power contained in the surface volume where the dielectric constant is modified by the analyte. This fraction can be increased by reducing the waveguide core layer thickness to a point where the mode becomes less confined and expands from the core,

raising the field magnitude at the interface. Furthermore, given the boundary condition that the normal component of the electric displacement vector $\mathbf{D} = \epsilon\mathbf{E}$ must be continuous across the interface, the field amplitude above the surface is increased by using a high-index-contrast waveguide and the TM mode, for which the dominant field component is polarized normal to the sensing surface. This results in a discontinuity of the normal component of $\mathbf{E}$ at the surface, which serves to enhance the field at the analyte side of the interface relative to that on the core side by a factor of the ratio of the respective dielectric constants. This enhancement factor thereby increases with index contrast.

For surface sensing, it is important that the evanescent tail of the waveguide mode is localized near the waveguide surface where the biomolecules are located and not extend too far into the analyte solution. The penetration depth of the waveguide mode into the cladding material of refractive index n_c is given by [20]

$$\gamma = \frac{\lambda}{2\pi\sqrt{N_{\text{eff}}^2 - n_c^2}}. \quad (2)$$

From (2), it follows that as the index contrast increases (which raises N_{eff}) the evanescent field becomes more localized near the waveguide surface.

Finally, it is important to maximize the proportion of the evanescent field on the analyte side of the waveguide core relative to that on the substrate (lower cladding) side. In most practical sensing applications where the lower cladding refractive index is greater than the analyte medium index, the sensitivity can be shown to be an increasing function of the waveguide symmetry [15]. The use of high refractive index core materials on low refractive index cladding materials increases the effective waveguide index symmetry, and the fraction of the optical power traveling through the lower cladding can be minimized.

3. WAVEGUIDE OPTIMIZATION AND SENSING CONFIGURATIONS

3.1. Waveguide design

The above arguments suggest that a high refractive index, thin waveguide core layer on a low refractive index lower cladding is an ideal choice for evanescent field sensors. Submicron dimension waveguides fabricated on the SOI material platform strongly meet these requirements. To optimize the SOI waveguide parameters and to examine how the thickness of the silicon waveguide core layer influences the evanescent field interaction with molecules located at the waveguide surface, we performed transfer-matrix-based waveguide simulations using a simple step-index, one-dimensional slab waveguide model. This model was used to calculate the effective index change induced by the adsorption of a thin biomolecular layer on the silicon waveguide core layer. An aqueous upper waveguide cladding and a silica lower cladding were assumed, and the signal wavelength was taken as 1550 nm. Working at this optical wavelength allows low cost and widely available optical test

FIGURE 2: (a) Calculated sensitivity constants for a SOI slab waveguide as a function of waveguide core thickness. (b) Calculated maximum $\partial N_{\text{eff}}/\partial d$ obtained for slab waveguides with varying refractive index contrast.

equipment developed for the telecommunications industry to be used. The calculations were performed for varying silicon waveguide thickness for both the transverse electric (TE) and TM polarization modes and are shown in Figure 2(a). As seen, the sensitivity has a clear maximum at a specific silicon waveguide thickness, which is 0.06 μm for TE polarization and 0.22 μm for TM polarization. Advantageously, the optimum silicon thickness value of $\sim 0.2 \mu\text{m}$ for the more sensitive TM polarization is a commonly used SOI wafer structure and is readily available from commercial suppliers.

Similar calculations were performed for waveguide systems with varying refractive index contrast (Δn) to confirm the sensitivity arguments discussed in Section 2. For consistency, all calculations were performed for a signal wavelength of 1550 nm, and an aqueous upper cladding and a silica lower cladding were assumed. The maximum achievable sensitivity is plotted in Figure 2(b). For the lower-index contrast systems ($\Delta n \leq 0.16$), a minor increase in sensitivity is obtained for the TE polarized mode compared with the TM mode. For larger index contrast platforms, the sensitivity is higher for the TM mode as expected from the

electric field boundary conditions discussed above. From Figure 2(b), the sensitivity advantage of using high-index-contrast waveguides such as SOI is clearly seen.

For a practical sensor, the waveguide should provide confinement in two dimensions and support only a single optical mode. The latter is required to obtain a well-defined spectral transmission function that can be easily analyzed. For single mode operation near the optimum silicon core thickness, silicon photonic wire waveguides, with the silicon etched fully down to the buried oxide, are a natural choice. The calculated TM mode field is shown in Figure 3 for the $0.26 \mu\text{m} \times 0.45 \mu\text{m}$ photonic wire waveguide used in our experiments. A large surface field magnitude is obtained along the top of the waveguide, which is localized within $\sim 120 \text{ nm}$ of the waveguide surface for strong interaction with surface bound molecules.

3.2. Fiber-to-waveguide optical coupling

A major challenge with the use of photonic wire waveguides is the difficulty of coupling light from an optical fiber with a mode size diameter of several μm to these waveguides with a submicron mode size. Due to the large mismatch, coupling losses exceeding 15 dB are common with such structures. To overcome this problem, we have developed optical mode transformers using inversely tapered waveguides covered with a 2- μm -thick SU-8 photoresist upper cladding layer. The sensor waveguide (nominal width of $0.45 \mu\text{m}$) is adiabatically tapered to a $0.15 \mu\text{m}$ wide coupling section at the device facet, as shown in Figure 4. At this narrow width, the mode becomes weakly confined in the core layer, thereby increasing its size in both the horizontal and vertical directions for greater overlap with the light emerging from the optical fiber. The mode transformers utilize a similar principle to the design reported in [21] and were developed for use with a lensed single-mode fiber. Using such a design, we have achieved less than 3 dB fiber-to-waveguide coupling loss, thus significantly improving the signal-to-noise ratio of our measurements compared to devices without mode transformers. Furthermore, since a large fraction of the optical field is traveling through the cladding materials in the tapered section near the device facet, the mode effective index is reduced. This minimizes reflection loss and Fabry-Pérot cavity effects, which cause instability of the fiber coupling due to mechanical vibration.

3.3. The Mach-Zehnder interferometer sensor

An evanescent field sensor relies on the perturbation of waveguide mode effective index that induces a phase shift of the propagating optical signal. This phase shift is not easily monitored directly and various interferometric waveguide circuits can be used to transform it to a more easily detectable intensity change. A simple and effective waveguide device for this purpose is the Mach-Zehnder interferometer (MZI), which is illustrated in Figure 5(a). In a typical configuration, light of wavelength λ (in vacuum) and intensity I_0 is split into two paths: a sensing path that is exposed to the analyte and an isolated reference path. These waveguide paths are

FIGURE 3: The electric field distribution for the TM polarization of a $0.26 \mu\text{m} \times 0.45 \mu\text{m}$ silicon photonic wire waveguide covered with an aqueous cladding. As seen, large electric field magnitude is obtained at the waveguide surface.

FIGURE 4: Schematic of the inversely tapered mode transformer used for improved fiber-to-waveguide coupling. The nominal $0.45 \mu\text{m}$ wide photonic wire sensor waveguide is laterally tapered down to $0.15 \mu\text{m}$ for adiabatic expansion of the waveguide mode near the device facet.

then recombined causing the signals from the two arms to interfere producing a phase dependent intensity I , to be observed at the MZI output. The transfer function of the MZI is

$$I = \frac{I_0}{2} \left[1 + \cos \left(\frac{2\pi}{\lambda} (N_{\text{eff},s} L_s - N_{\text{eff},r} L_r) \right) \right], \quad (3)$$

where $N_{\text{eff},s}$ and L_s are the effective refractive index and length of the sensing waveguide, and $N_{\text{eff},r}$ and L_r are the effective refractive index and length of the reference waveguide. The phase difference between the two arms is given by the argument of the cosine function in (3). By differentiating this

quantity, the sensitivity of the phase ϕ to homogeneous and surface sensing can be expressed as

$$\begin{aligned} \frac{\partial \phi}{\partial n_c} &= \frac{2\pi L_s}{\lambda} \frac{\partial N_{\text{eff},s}}{\partial n_c} \quad (\text{homogeneous sensing}), \\ \frac{\partial \phi}{\partial d} &= \frac{2\pi L_s}{\lambda} \frac{\partial N_{\text{eff},s}}{\partial d} \quad (\text{surface sensing}). \end{aligned} \quad (4)$$

It is seen that the phase sensitivity scales with sensing waveguide length L_s and with the waveguide sensitivity constants: $\partial N_{\text{eff}}/\partial n_c$ or $\partial N_{\text{eff}}/\partial d$. MZI devices provide important practical advantages over other waveguide circuit designs. Since the intensity change is monitored, they can be used with a low cost, fixed wavelength laser source and by tracking the sinusoidal output variations they can function over a large dynamic range of surface loading conditions. Furthermore, by balancing the optical path lengths of the sensor and reference arms the MZI can be designed to function independently of temperature and signal wavelength. This reduces the requirements for temperature stability of the chip and wavelength stability of the laser source, thereby reducing potential instrumentation costs associated with this technology.

3.4. The ring resonator sensor

The high lateral index contrast provided by silicon photonic wire waveguides allows waveguide bends to be fabricated with a radius of curvature of only a few micrometers with little excess loss. This has led to the development of ultracompact ring resonator structures, which have been used for many applications, including optical add/drop filtering [22], sensing [18], optical logic [23], nonlinear optics [24], and optical modulation [25].

In the simplest form, a ring resonator contains a straight bus waveguide evanescently coupled to a ring cavity using a directional coupler, as illustrated in Figure 5(b). Light of specific wavelengths couples from the bus waveguide to the ring waveguide and circulates, making many round trips and building up intensity within the ring cavity. At these resonant wavelengths, a sharp dip is observed in the transmission spectrum of the bus waveguide. The sharpness of the resonance dip is described by the quality factor Q of the ring [26]:

$$Q \equiv \frac{\lambda}{\Delta\lambda_{\text{FWHM}}} \cong \frac{\pi n_g L}{\lambda} \frac{\sqrt{\alpha t}}{1 - \alpha t}, \quad (5)$$

where $\Delta\lambda_{\text{FWHM}}$ is the full width at half maximum of the resonance, L is the ring cavity length, n_g is the ring waveguide group index, t is the through coupling coefficient between the bus and ring waveguides, and α is the total resonator loss including the coupler loss and the round trip propagation loss in the ring.

An undesirable characteristic of conventional ring resonators with a directional coupler is the inherent wavelength dependence of the coupling between the bus and ring waveguides. This causes a slowly varying modulation of the resonance depths in the output spectrum, limiting the

FIGURE 5: (a) Schematic of an MZI evanescent field sensor. The input waveguide is split into two paths where one is exposed to the analyte and the other is isolated from the analyte. The paths are recombined and the signals interfere causing a sinusoidal phase dependent intensity signal to be transmitted through the output waveguide. (b) A conventional ring resonator design employing a directional coupler between the straight bus waveguide and the ring cavity. The dashed lines indicate the positions of the sensor windows where the upper SU-8 isolation layer is removed.

useable wavelength range over which a uniform transmission spectrum is obtained. To overcome this limitation, we have designed a new ring resonator structure in SOI that utilizes a multimode-interference coupler (MMI) between the bus and ring waveguides [26]. The MMI significantly reduces the wavelength dependence of the coupling, producing a more uniform resonance spectrum over a broad range of wavelengths. In addition, the MMI structure improves the fabrication tolerance by eliminating the narrow gap of the directional coupler.

As molecules bind to the ring surface (or if the bulk refractive index of the analyte solution changes) a phase shift of the optical mode is induced, resulting in a wavelength shift of the resonances. This wavelength shift scales with the number of adsorbed molecules to the ring waveguide (or with the refractive index change of a bulk solution) and is typically the parameter being measured in this type of sensor. Alternatively, it is also possible to track the intensity change at a fixed wavelength over a single resonance peak. In this case, the dynamic range is limited and the sensor can only be used to track comparatively low analyte concentrations or low refractive index changes of bulk solutions since the peak position may shift away from the input wavelength. However, active tuning of the resonance through temperature or other means may provide a solution to this tradeoff at the price of increased complexity.

Ring resonators have very sharp resonances and may provide higher sensitivity compared to other waveguide circuit configurations such as the MZI. Since light at the resonance wavelengths circulates around the ring cavity, it has numerous interactions with the same molecules amplifying the measured response. They also offer a more compact geometry desirable for the fabrication of high-density sensor arrays. However, when the resonance wavelengths are

monitored, ring resonators require either a tunable laser source or a broadband optical source and a means to monitor the output wavelengths. In either case, this increases the potential instrumentation costs for this technology. Ring resonators also do not provide for internal referencing for temperature or wavelength drifts and therefore require either precise temperature control of the sensor and optical source, or they require a second reference device.

Due to the different characteristics of the ring resonator and the MZI, we are currently developing both types of sensors where we believe that the optimal circuit choice will depend on the sensitivity and cost requirements of the particular application.

4. DESIGN AND FABRICATION

MZI sensors and ring resonators were designed having $0.26\ \mu\text{m} \times 0.45\ \mu\text{m}$ photonic wire sensing waveguides. The waveguide height was chosen to be near the optimum sensitivity point of Figure 2(a) for TM polarization. The waveguide width was designed to be narrow enough to provide single mode operation, while being wide enough to permit sharp bends of the waveguide with a radius of curvature of a few microns. Optical mode transformers having a $0.26\ \mu\text{m} \times 0.15\ \mu\text{m}$ cross section were integrated at both waveguide facets for improved input and output coupling. Our early designs used conventional waveguide layout geometries for the ring and MZI sensors and are shown in Figure 6. We have demonstrated sensitive evanescent field sensors using these devices, capable of detecting less than 1% of a monolayer of streptavidin molecules adsorbed to the biotinylated waveguide surface [18].

Here, we will focus on our more recent sensor designs that further exploit the high lateral index contrast of the silicon photonic wire waveguide, which allows very sharp waveguide bends to be achieved. As mentioned above, the phase sensitivity of evanescent field sensors is determined by both the intrinsic waveguide sensitivity constants ($\partial N_{\text{eff}}/\partial n_c$ or $\partial N_{\text{eff}}/\partial d$) and the waveguide length. By increasing the waveguide length, a larger number of molecules will be captured on the waveguide surface for a given analyte concentration and interaction time, thereby inducing a larger phase shift (see (4)). However, long straight waveguides are not desirable for two-dimensional arraying purposes and are not suited for common functionalization techniques using commercial spotting apparatus. To address these issues, we have designed coiled waveguide sensing structures as shown in Figure 7. For the MZI devices, several mm of sensing path length have been designed in a circular area of $150\ \mu\text{m}$ diameter, comparable to the drop size employed by modern spotting tools. Spiral ring cavities with 1.3 mm length have also been designed with similar dimensions. The additional length over conventional circular/oval cavities leads to improvement of the quality factor of the resonance allowing increased measurement accuracy of the resonance wavelengths. Both the spiral MZI and ring designs offer the high sensitivity of a long waveguide without sacrificing

(a)

(b)

FIGURE 6: Scanning electron microscope images of our early sensor designs employing silicon photonic wire waveguides arranged in: (a) a conventional MZI structure and (b) a ring cavity containing a MMI coupler.

size, and their increased surface areas provide more efficient molecular capture.

The devices in Figure 7 were fabricated on SOI wafers having a $0.26\text{-}\mu\text{m}$ -thick silicon layer on a $2\text{-}\mu\text{m}$ -thick buried oxide layer. Waveguides were patterned by electron beam lithography and etched in an inductively coupled plasma reactive ion etching system. To isolate the optical mode of the nonsensing portions of the device from the environment, a $2\text{-}\mu\text{m}$ -thick layer of SU-8 photoresist was spun on the wafer and windows were opened over the active sensing area using optical contact lithography. The devices were then baked to cure the SU-8 film. Finally, samples were exposed to an O_2 plasma to remove any residual photoresist in the sensor windows as well as to increase the thickness of the native oxide layer (used later for functionalization). The sensors were then diced into individual chips and some were then used for homogeneous solution sensing experiments as described in Section 5.

For surface sensing experiments, the sensors were functionalized by immobilizing biotin on the chip surface in order to monitor the affinity reaction between the vitamin biotin and the protein streptavidin. The oxide on the silicon waveguide surface, measured to be $2.7 \pm 0.2\ \text{nm}$ thick by ellipsometry, was left intact so that standard glass-based functionalization procedures to attach biotin to the surface

FIGURE 7: Sensor designs with double spiral waveguide geometries providing increased sensing path length while maintaining a compact sensor footprint. (a) MZI sensor with 2 mm sensing path length contained within a circular 150 μm diameter area. (b) Spiral cavity ring resonator design with 1.3 mm sensor length contained within a circular 110 μm diameter area.

could be used. The functionalization procedure began with a cleaning step where the sample was rinsed with ethanol and then cleaned and activated with nitric acid for 5 minutes. The samples were then rinsed with deionized water and ethanol and dried under nitrogen. Surfaces were then silanized with 3-aminopropyltriethoxysilane (APTES) vapor. The surfaces were rinsed with ethanol, dried and biotinylated by coating the surface in a solution of 1 mg/mL of N-hydroxysuccinimide (NHS) activated biotin in dimethylformamide (DMF) for 1 hour, followed by rinsing with DMF and ethanol. The incremental thickness change induced by adsorption of streptavidin to the sensor surface was assessed using a functionalized silicon wafer, which was immersed in a streptavidin solution for 1 hour followed by rinsing with phosphate buffered saline (PBS). Based on ellipsometry measurements, we estimate a surface mass coverage of 2.2 ng/mm² of streptavidin, following Elwing [27]. The above process was performed on the entire sensing chip. Future experiments will attempt to functionalize individual sensor elements using a microarray spotter for multianalyte detection.

Analyte was delivered to the sensor through a microfluidic component fabricated in polydimethylsiloxane (PDMS).

FIGURE 8: MZI sensor response to homogeneous solution refractive index change. The calculated phase shift values are shown by the dotted line.

Channels with a width of 200 μm were defined in the PDMS by using a surface relief mold fabricated using 50 μm thick SU-8 resist tracks defined on a silicon wafer. The PDMS component was subsequently aligned and fixed to the sensor chip, and a syringe pump was used to drive solutions through the microfluidic channels.

5. EXPERIMENTAL DEMONSTRATION

Using a tapered optical input fiber, a tunable laser source, and an InGaAs photodetector, the propagation loss of the fabricated waveguides was assessed by the Fabry-Pérot technique. Several waveguides were measured with loss values varying between 0.3-0.4 dB/mm. The optical coupling loss from a tapered optical fiber was found to be less than 3 dB per facet using the inversely tapered mode converters.

The response of an MZI sensor with a 2-mm-long spiral sensing arm, to homogeneous refractive index change of the analyte solution, was then evaluated by passing sucrose solutions of varying concentration through the microfluidic channel. The solutions were drawn over the exposed sensor arm at a rate of 2 mL/hour, and the transmission of the sensor was measured for each solution. The corresponding phase shift values are shown in Figure 8 as a function of sucrose solution refractive index change (with respect to water). At a signal wavelength of 1550 nm, these phase shift values correspond to a large intrinsic sensitivity constant of $\partial N_{\text{eff}}/\partial n_c = 0.35$, which is only approximately three times less than that achievable using a free-space beam passing through a bulk solution cell. These results agree to within 5% of the calculated phase shift values from the effective indices obtained in mode expansion-based waveguide simulations.

Using the biotin-streptavidin binding system, the response of an MZI with a 2-mm-long spiral sensing arm and a ring resonator sensor with a 1.3-mm-spiral cavity length to the adsorption of a thin protein monolayer was then examined. The intensity change at the MZI output was

FIGURE 9: Surface adsorption measurements using the sensors to monitor the affinity reaction between streptavidin and the biotinylated waveguide surface. (a) Measured shift in the resonance wavelength for a ~ 1.3 -mm-long double spiral ring resonator sensor. (b) Measured intensity change (grey) and phase shift (black) of an MZI sensor with a 2-mm-long double spiral sensing arm.

monitored at a constant signal wavelength of ~ 1550 nm, whereas the wavelength shift of the ring resonator resonance was monitored during the experiment by sweeping a tunable laser and recording the transmission spectrum. The results of the measurements are shown in Figure 9(a) for the ring resonator sensor and Figure 9(b) for the MZI sensor.

For the ring resonator, the first analyte solution consisted of $0.1 \mu\text{g/mL}$ of streptavidin in PBS. It was passed through the microfluidic channel over the ring cavity, and the transmission spectrum of the ring was recorded every 5 seconds over a 3-nm-wavelength range. Molecular binding is observed by the wavelength shift in the ring resonator response as shown in Figure 9(a). The streptavidin concentration was then increased to $1 \mu\text{g/mL}$ and then to $10 \mu\text{g/mL}$, where successively increased binding rates are observed. For the MZI device, a constant wavelength laser source was used, and the transmitted power was recorded as solutions were drawn over the sensing arm. The results of the experiment are shown in Figure 9(b) for both the raw transmitted intensity signal and the corresponding phase change. The experiment began by flowing PBS solution through the microfluidic

channel for 60 seconds. The solution was then switched to a $10 \mu\text{g/mL}$ streptavidin solution, followed by a PBS solution to rinse nonspecifically bound molecules from the waveguide surface. A typical high-affinity binding curve is obtained. Taking the RMS noise levels in Figure 9 as the minimum detectable phase change or wavelength shift, we estimate that the MZI sensor is capable of detecting $\sim 0.3\%$ of a protein monolayer, while the ring resonator demonstrates a resolution of $\sim 0.2\%$ of a monolayer. These values correspond to a minimum detectable surface coverage of $\sim 6 \text{ pg/mm}^2$ for the MZI sensor and $\sim 4 \text{ pg/mm}^2$ for the ring resonator, which are comparable to the sensitivities reported for state-of-the-art SPR sensors. We have established that the sensor performance is currently limited by intensity noise resulting from Fabry-Pérot cavity effects between the fiber tip and the device facet. These effects are lessened with the inverse tapered mode transformers and can be further reduced by depositing an antireflection coating on the input waveguide facet. Furthermore, for applications where a larger sensor area is permitted the waveguide length can be increased for improved performance. With further optimization of the sensor design and with improvement of the experimental setup to reduce the measurement noise, we expect that minimum detectable surface coverage values well below 1 pg/mm^2 are attainable.

6. SUMMARY

We have reviewed our work developing biological sensors based on silicon photonic wire waveguides arranged in an MZI and ring resonator configuration. We have shown that high-index-contrast SOI waveguides provide a large and localized evanescent field for the TM polarized waveguide mode, resulting in high surface sensitivity. We have also used the high lateral index contrast of silicon photonic wires to make densely coiled sensor waveguide structures that provide the advantages of long sensing path length without compromising the sensor footprint. Due to their compact size, these sensors are compatible with spotter functionalization, microfluidic sample delivery, and the fabrication of two-dimensional arrays. Our future work will focus on the development of an evanescent field-based microarray biochip for multianalyte monitoring that exploits the benefits offered by this technology.

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