

Guided mode biosensor based on grating coupled porous silicon waveguide

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Abstract: Porous silicon waveguide biosensors that utilize grating couplers etched directly into porous silicon are demonstrated for improved molecular detection capabilities. Molecules are infiltrated through the grating couplers into the waveguide where they can interact with a guided waveguide mode. Hybridization of nucleic acids inside the waveguide is shown to significantly perturb the wave vector of the guided mode and is detected through angle-resolved reflectance measurements. A detection sensitivity of 7.3°/mM is demonstrated with selectivity better than 6:1 compared to mismatched sequences. Experimental results are in good agreement with calculations based on rigorous coupled wave analysis. Use of the all-porous silicon grating-coupled waveguide allows improved interaction of the optical field with surface-bound molecules compared to evanescent wave-based biosensors.

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1. Introduction

For more than two decades, gratings have been combined with waveguide structures to form biosensors because of their outstanding integrability and relatively high sensitivity detection of analytes [1–9]. Although these grating-coupled sensors have been demonstrated using a variety of different materials and different configurations, they all realize label-free detection by means of affecting the evanescent tail of a waveguide mode near the interface. When target analytes bind to surface-bound probe molecules, the effective refractive index of the media surrounding the waveguide is changed, resulting in a perturbation to the original waveguide mode. By monitoring this perturbation, molecular binding can be detected. The key disadvantage of this evanescent wave-based detection method is that typically >80% of the total optical power is contained within the waveguide [10] and does not contribute to the surface attachment sensing; the evanescent wave that interacts with molecules is an exponentially decaying field that only exists within a few hundred nanometers, or even tens of nanometers, of the waveguide interface [11]. Therefore, detection of small molecule binding, which causes only a small perturbation to the field distribution, is challenging for evanescent wave sensors and results in low detection sensitivity. The formation of silicon wire waveguides that enable a larger overlap of the evanescent field with molecules allow for higher sensitivity detection but at the cost of more challenging optical coupling requirements [12].

In this paper, we combine gratings with porous silicon (PSi) waveguides to form compact biosensors that can detect molecular binding events in a straightforward manner based on guided mode perturbations. PSi is a crystalline form of silicon permeated with nanoscale void spaces. Hence, molecules can be infiltrated directly into the core of the waveguide where they interact with a guided mode. The large internal surface area of PSi (up to 800 m²/g [13]) allows immobilization of more probe molecules compared to a flat surface, which increases the likelihood of capturing low concentration target molecules in a complex solution.

In section 2, we first present the basic theory of grating coupled waveguides and compare the field distribution and potential sensing capabilities of three different configurations. Next, in section 3, fabrication, surface functionalization, and experimental measurement procedures are described. Finally, experimental results and analysis of grating-coupled PSi waveguides as DNA sensors are shown in section 4.

2. Theory

The fundamental geometry of the grating coupled waveguide is depicted in Fig. 1(a). The basic theory of operation, which is described and demonstrated in Ref. [14], can be summarized as follows. An incoming light beam I_0 is reflected into several diffraction orders by the grating. The waveguide structure, which consists of a high-index waveguide layer surrounded by a lower-index substrate and air, can support only a discrete number of guided modes [15]. When the wave vector of one diffracted order beam matches that of one guided mode, coupling occurs. The coupling condition can be calculated from the well-known grating equation

$$n_{\text{eff}} = n_c \cdot \sin \theta + m \cdot \frac{\lambda_0}{\Lambda} \quad (1)$$

where n_{eff} is the effective refractive index of the guided mode, n_c is the refractive index of the cover layer (air in the schematic shown in Fig. 1(a)), θ is the coupling angle, m is the diffraction order, λ_0 is the wavelength in free space, and Λ is the grating period.

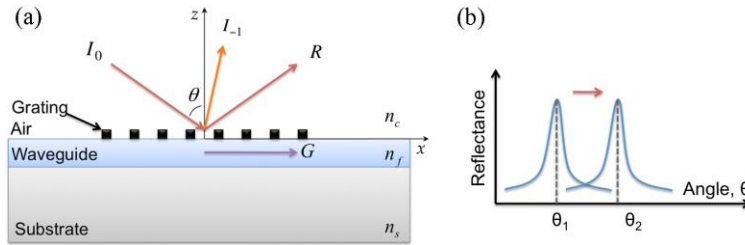


Fig. 1. (a) Schematic of a grating coupler on a waveguide. The incident beam (I_0), zeroth order reflected beam (R), minus first-order diffracted beam (I_{-1}), and the guided mode (G) are shown. (b) The reflected light intensity peaks at the angle corresponding to light coupling into the waveguide and depends on the effective index of the waveguide mode such that biomolecule infiltration shifts the angular position of the reflection peak.

When the diffracted light is coupled into the waveguide, the zeroth-order transmitted intensity will drop. In reflection, however, a narrow peak in intensity is observed. This is due to the principle of reversibility, which states that if light can be coupled into the waveguide then it can also be coupled out of it [16]. The out-coupled light is coincident with the reflected beam, which increases the intensity of the reflected light at that given angle. If biomolecules are attached inside the waveguide and change the effective refractive index of the guided mode, the coupling angle θ will be changed according to Eq. (1). The angular position of the reflection peak will shift, accordingly, as illustrated in the Fig. 1(b). Therefore, biomolecule interactions can be monitored quantitatively by measuring the magnitude of this angular shift.

In order to evaluate the potential sensing capabilities of grating-coupled waveguides and better understand the advantages of the PSi guided mode sensor, we examine three different configurations of grating-coupled waveguides, as illustrated in Fig. 2. First, for baseline comparison, a traditional silicon-on-insulator (SOI) waveguide with SiO_2 grating, following [8], is considered. Next, PSi waveguides with either photoresist gratings, following [14], or lithographically etched PSi gratings are considered. PSi waveguides with lithographically etched PSi gratings are proposed for the first time here and enable the entire internal surface area of the PSi films to be accessible for molecular infiltration. We refer to these structures as all-PSi grating-coupled waveguides. The dimensions of the all-PSi grating-coupled waveguides were chosen such that the total PSi thickness was similar to that of the photoresist grating-coupled PSi waveguide.

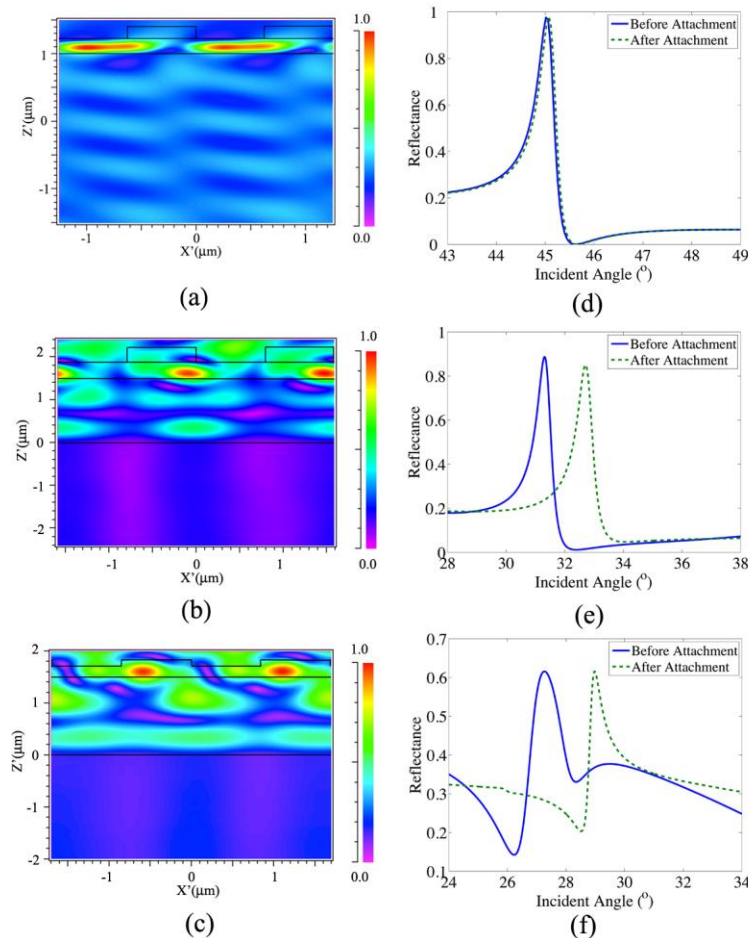


Fig. 2. RCWA calculations showing (a) magnetic field distribution of the SOI guided mode resonance sensor (thickness of gratings $h_g = 180\text{nm}$; grating period $\Lambda = 1240\text{nm}$; refractive index of gratings $n_g = 1.45$; thickness of waveguide layer $h_w = 220\text{nm}$; refractive index of waveguide layer $n_w = 3.476$; thickness of substrate $h_s = 3000\text{nm}$; refractive index of substrate $n_s = 1.45$; incident light wavelength $\lambda = 1531\text{nm}$) (b) electric field distribution of the photoresist grating-coupled PSi waveguide ($h_g = 380\text{nm}$; $\Lambda = 1590\text{nm}$; $n_g = 1.54$; $h_w = 340\text{nm}$; $n_w = 1.80$; $h_s = 1500\text{nm}$; $n_s = 1.21$; $\lambda = 1550\text{nm}$); and (c) electric field distribution of the all-PSi grating waveguide structure ($h_g = 135\text{nm}$; $\Lambda = 1685\text{nm}$; $n_g = 1.80$; $h_w = 190\text{nm}$; $n_w = 1.80$; $h_s = 1500\text{nm}$; $n_s = 1.21$; $\lambda = 1550\text{nm}$). Reflectance spectra of (d) SOI guided mode resonance sensor, (e) photoresist grating-coupled PSi waveguide, and (f) all-PSi grating-coupled waveguide structure before and after attaching a 0.8 nm thick monolayer of biomolecules.

Rigorous coupled-wave analysis (RCWA [17]) calculations are performed to determine the field distributions in each of the grating-coupled waveguide structures and to evaluate the potential sensing capabilities of each design. For the SOI waveguide, molecules can only be attached to the top surface of the grating-waveguide structure while for the PSi waveguides, biomolecules can be additionally attached to the pore walls inside the waveguide. Note that for the photoresist gratings on PSi waveguides, only half of the available surface area of the PSi film is accessible for molecular infiltration. As shown in Fig. 2(a)–2(c), only 1.67% of the transverse magnetic (TM) field power distribution is localized at the top surface (within 180nm of the waveguide surface in the cover region) of the SOI grating-coupled waveguide where molecules can be attached. In contrast, for all-PSi grating-coupled waveguides, 54.07% of the transverse electric (TE) field power distribution is localized within the waveguide and at the top surface where molecules can be attached. For PSi waveguides with photoresist

gratings, the surface area accessible for molecular infiltration is reduced and only 23.27% of the TE field power distribution is localized where molecules will be present. Note also that the fraction of field power in the grating region is increased for the PSi waveguides with photoresist gratings (16.03%) compared to those with etched PSi gratings (3.72%).

Figure 2(d)–2(f) show the theoretical reflectance spectra, based on the RCWA calculations, of the three grating-coupled waveguide structures before and after attaching a 0.8 nm thick biomolecule monolayer ($n_{bio} = 1.45$; e.g., 3-aminopropyltriethoxy-silane). As expected, the magnitude of the resonance shift for each structure directly relates to the fraction of field overlap with molecules [18]. The all-PSi grating waveguide sensor exhibits a 1.72° resonance angle shift, the photoresist grating-coupled PSi waveguide sensor exhibits a 1.4° shift, and the SOI guided mode resonance sensor exhibits a 0.05° shift due to molecular attachment. We note that the 0.05° resonance angle shift for the SOI guided mode resonance sensor corresponds to an equivalent refractive index change as a resonance wavelength shift of approximately 0.76 nm (for a fixed angle of 45° , a grating period of 1240 nm, and a resonance wavelength near 1530 nm) [19], which is consistent with the simulations and data presented in [8]. Considering the 0.8 nm biomolecules size, the sensitivity of the SOI guided mode resonance sensor is ≈ 1 nm/nm. Similarly, the detection sensitivities of the all-PSi grating waveguide sensor and the photoresist grating-coupled PSi waveguide sensor are calculated to be ≈ 56 nm/nm and ≈ 41 nm/nm, respectively. Clearly, the additional available surface area in the PSi drastically improves the grating-coupled waveguide sensor performance for small molecule detection. Use of the etched PSi grating instead of the photoresist grating further increases the available active sensing surface area, and will be investigated in detail in the following sections.

3. Method

3.1 Fabrication

The two-layer PSi waveguide structures used in this work were fabricated by electrochemical etching of p + (0.01 Ω ·cm) silicon in 15% ethanoic hydrofluoric acid, similar to what we have reported previously [14]. The top PSi layer that constitutes the waveguide layer with low porosity (high refractive index) was etched at 5 mA/cm² for 62 sec. The bottom PSi layer that constitutes the substrate layer with high porosity (low refractive index) was subsequently etched at 48 mA/cm² for 53 sec. The PSi waveguide was then soaked in 1.5 mmol·L⁻¹ KOH solution for 30 minutes to widen the pores to promote biomolecule penetration, followed by thermal oxidation at 500°C for 5 min.

PSi grating couplers were fabricated on the PSi waveguide by electron beam lithography and reactive ion etching (RIE). A 300 nm film of PMMA 950 photoresist was spun onto the PSi waveguide and exposed by a Raith eLINE electron beam lithography tool to form a diffraction grating with a grating period of approximately 1700 nm. After development, the PSi waveguide with PMMA gratings was reactive ion etched (Trion Technology) with 30 sccm SF₆ flow under 100 W RF power and 30 mTorr chamber pressure for 60 seconds. The PMMA gratings served as the etch mask for the formation of the PSi gratings. After the pattern transfer by RIE was completed, residual PMMA was removed by acetone. Figure 3 shows a cross-sectional scanning electron microscopy (SEM) image of the all-PSi grating-coupled waveguide. Based on this SEM image and other SEM images, it was determined that the PSi grating height is 132 nm, the period is 1682 nm, and the air fill factor is 49%. Moreover, the waveguide and substrate layers were determined to have thicknesses of 190 nm and 1507 nm, respectively. Based on reflectance spectra measurements and analysis [14], the refractive indices of the waveguide and substrate layers were determined to be 1.80 and 1.21, respectively.

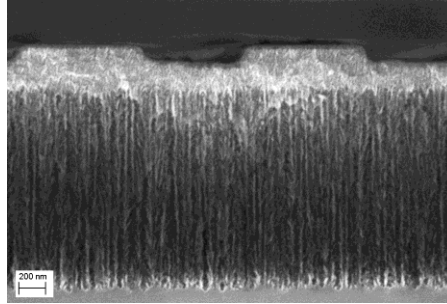


Fig. 3. SEM image of the all-PSi gratings and the waveguide (cross-section).

In order to realize experimental comparison of the performance of the PSi waveguide sensors with etched PSi gratings and photoresist gratings, we also fabricated photoresist gratings following the methods reported in [14, 20]. The PSi waveguide structure was made using the same procedure mentioned at the beginning of this section. Then, an approximately 400 nm thick film of ZEP 520A photoresist (Zeon Corp.) was spun onto the PSi waveguide and exposed by a JEOL-9300FS electron beam lithography tool to form a diffraction grating with a grating period of 1655 nm. Figure 4(a) shows a cross-sectional SEM image of the photoresist gratings on a PSi waveguide, revealing a grating height of 380 nm, 340 nm thick waveguide layer, and 1485 nm thick substrate layer. An SOI waveguide with SiO₂ gratings was also fabricated to enable experimental comparison of sensor performance for DNA hybridization. First, 180 nm of amorphous SiO₂ was deposited on an SOI substrate (220 nm thick silicon and 2 μ m thick buried oxide layer) by plasma enhanced chemical vapor deposition using a Trion Orion II System. Similar to the method to fabricate photoresist gratings on the PSi waveguide, ZEP 520A photoresist was spun onto the amorphous SiO₂ coated SOI wafer and exposed by electron beam lithography (Raith eLINE) to form gratings with 1240 nm period. After serving as a RIE mask to etch the SiO₂ underneath, the photoresist was then removed by Remover PG (MicroChem Corp.). Figure 4(b) shows a cross sectional SEM image of the SOI waveguide with SiO₂ gratings, revealing a grating height of 180 nm.

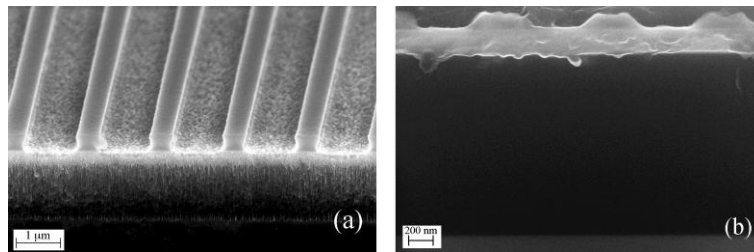


Fig. 4. SEM images of (a) photoresist gratings on top of a PSi waveguide and (b) SOI waveguide with SiO₂ gratings. The image in (a) was taken at an angle of 30° to the planar surface. The apparent non-planarity of the interface between the silicon waveguide and SiO₂ substrate layer in (b) is an artifact due to the sample cleavage.

3.2 Surface functionalization

Complete details of the process used to functionalize the grating-coupled waveguides in this work can be found in Ref. [21]. Briefly, in order to attach probe DNA oligos to an oxidized PSi waveguide sample, an organofunctional silane, 3-aminopropyltriethoxysilane (3-APTES, ACROS), which has a terminal amine group, was used to modify the silica surface. The oxidized PSi waveguide sample was soaked in 4% 3-APTES solution for 20 minutes. Then it was rinsed with deionized water and baked at 100°C for 10 minutes to promote cross-linking and remove excess solvent. The silanized PSi waveguide was incubated in 2.5 mg·mL⁻¹ Sulfo-succinimidyl 4-[N-maleimidomethyl]cyclohexane-1-carboxylate (Sulfo-SMCC, Pierce)

in 4-(2-hydroxyethyl)-1-piperazineethanesulfonic acid (HEPES) buffer solution for 2 hours, followed by a 1 hour soak in HEPES buffer, rinsing with deionized water, and drying with nitrogen gas. 16-base thiol modified probe DNA (5'-TAG CTA TGG TCC TCG T-3', 3' Thiol C3, Eurofins MWG Operon) in HEPES buffer was mixed 1:1 by volume with TCEP (Pierce) in water and ethanol for 30 minutes, and then directly infiltrated into the functionalized PSi waveguide sample. After 1 hour incubation at 37°C, the sample was soaked in HEPES buffer for 20 minutes at the same temperature, rinsed with deionized water, and dried with nitrogen gas to remove any remaining unattached molecules. Complementary peptide nucleic acid (PNA, ACG AGG ACC ATA GCT A, BioSynthesis) is chosen as the target molecule. Since the PNA contains no charged groups, it has been shown that the binding between PNA/DNA strands is stronger than between DNA/DNA strands due to the lack of electrostatic repulsion [22]. Complementary PNA in HEPES buffer was dropped on the sample and incubated at 37°C for 1 hour. Then the sample was soaked in HEPES buffer for 20 minutes to remove non-hybridized oligos, rinsed with deionized water, and dried with nitrogen gas. The concentrations of the DNA and PNA molecules were varied for different experiments and are noted accordingly in the results and discussion section. The 1 hour incubation times used for DNA and PNA incubation were not optimized; prior studies of molecular infiltration into porous silicon sensors suggest that shorter times may be feasible [23, 24].

3.3 Experimental setup

A Metricon Model 2010/M Prism Coupler (Metricon Corp., USA) was used in the non-contact, variable-angle monochromatic fringe observation (VAMFO) mode (i.e., without a prism) to monitor the reflectance of the grating-coupled waveguides. A 1550 nm diode laser was used as the light source. A schematic of the measurement configuration for the grating coupled waveguides is shown in Fig. 5. Light from the laser is incident on the waveguide at variable angle by rotating the stage, and the reflected light intensity is measured by an InGaAs photodetector.

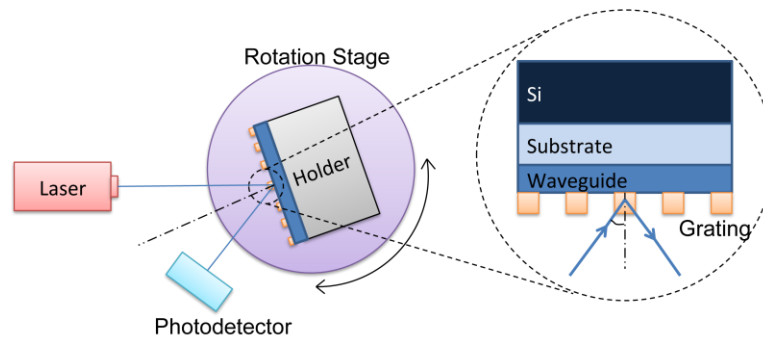


Fig. 5. Schematic of the measurement configuration for the grating coupled waveguides.

4. Results and discussion

4.1 All-PSi grating-coupled waveguide biosensor

Using the setup shown in Fig. 5, angular reflectance measurements of the all-PSi grating-coupled waveguide were taken after each functionalization step in order to confirm the molecular attachments described in section 3.2. The reflectance spectra before and after the 3-APTES attachment are shown in Fig. 6(a). The waveguide resonance angle is shifted to a higher angle (1.77° shift) due to the increase in refractive index of the PSi layers that results from the 3-APTES attachment on the pore walls. In order to verify monolayer attachment of 3-APTES (0.8 nm size, $n_{3\text{-APTES}} = 1.45$), RCWA simulations were performed using the experimentally determined dimensions of the all-PSi grating-coupled waveguide. In Fig. 6(b), we plot the calculated reflectance before and after attachment of a monolayer of 3-APTES

molecules, which indicates a resonance shift of 1.80° . Thus, the calculated angular reflectance spectra exhibit good agreement to the experimental data, and we can assume nearly complete monolayer coverage of the small 3-APTES molecules in the pores. Based on the RCWA calculations, we find that the refractive index changes caused by 3-APTES attachment are $\Delta n_{\text{waveguide}} = 0.0427$ and $\Delta n_{\text{substrate}} = 0.0509$. Hence, the detection sensitivity of the all-PSi grating-coupled waveguide for small molecules is approximately $42^\circ/\text{RIU}$, corresponding to $1090 \text{ nm}/\text{RIU}$ for wavelength interrogation measurement. The detection limit of the sensor depends ultimately on the measurement instrument resolution. Given the Metricon prism coupler angular resolution of 0.0075° in the VAMFO mode or considering a spectral measurement instrument with a modest resolution of 0.1 nm , the anticipated detection limit of the sensor is on the order of 10^{-4} RIU .

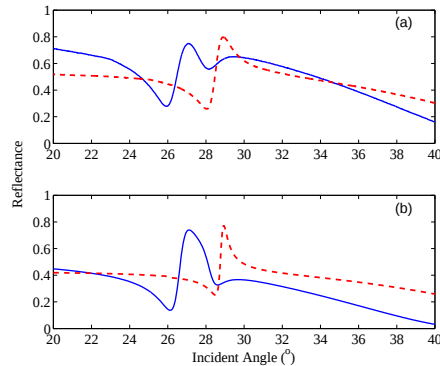


Fig. 6. (a) Measured and (b) calculated angle-resolved reflectance spectra at 1550 nm (TE) for all-PSi grating-coupled waveguide after oxidation (solid line) and after 3-APTES attachment (dashed line).

Figure 7(a) shows the reflectance spectra of the all-PSi grating-coupled waveguide after the sequential attachment of 3-APTES, Sulfo-SMCC, 16-mer probe DNA ($50 \mu\text{mol}\cdot\text{L}^{-1}$), and 16-mer complementary target PNA ($50 \mu\text{mol}\cdot\text{L}^{-1}$). The measured resonance shifts are stable over time durations of several minutes, suggesting that there is no liquid remaining trapped in the pores after each molecular infiltration step that can evaporate during the duration of the experiment. Note that attachment of Sulfo-SMCC leads to a larger resonance peak shift than 3-APTES attachment, which is consistent with their molecular size (3-APTES monolayer: 0.8 nm [25]; Sulfo-SMCC monolayer: 1.27 nm [26]). The resonance shift due to probe DNA attachment is comparatively small with respect to its molecule size ($\sim 3.5 \text{ nm}$ in length), suggesting that there is a relatively low probe density in the PSi waveguide. Based on the magnitude of the experimentally measured resonance shift (0.42°), RCWA calculations suggest that the probe DNA coverage on the pore walls is approximately 10%. This low coverage is attributed primarily to the size-dependent infiltration efficiency of small molecules into the small pore diameters ($\sim 20 \text{ nm}$) [21], as well as a possible contribution of charge-based exclusion of the DNA molecules. When exposed to the complementary PNA sequence, the majority of the probe DNA molecules were hybridized. Based on the magnitude of the experimentally measured resonance shift after PNA attachment (0.39°), we estimate a hybridization rate of $\sim 90\%$.

In order to confirm the selectivity of the all-PSi grating-coupled waveguide sensor, a mismatched sequence of PNA (ACG AGG ACC ATA GCT A) and HEPES buffer solution were separately exposed to similarly prepared all-PSi grating-coupled waveguides. As expected, and as shown in Fig. 7(b), the resulting shifts of the waveguide resonance angle for mismatch PNA (0.07°) and HEPES buffer (0.015°) were substantially less than that for the complementary PNA sequence.

Finally, in order to establish an approximate sensitivity for the all-PSi grating-coupled waveguide, several hybridization experiments were performed with fixed probe DNA concentration of $100 \mu\text{mol}\cdot\text{L}^{-1}$ and variable complementary PNA concentration. As shown in Fig. 8, the linearity of the resonance angle shift as a function of PNA concentration suggests a detection sensitivity of approximately $7.3^\circ/\text{mM}$. Note that the data point at $0 \mu\text{M}$ PNA concentration corresponds to the resulting resonance shift from exposure to the HEPES buffer alone. Given the angular resolution of the Metricon tool in the VAMFO mode, the PNA detection limit is expected to be on the order of $1 \mu\text{M}$.

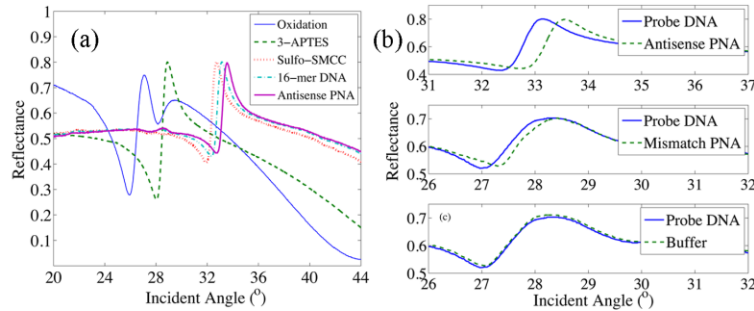


Fig. 7. (a) Reflectance spectra of all-PSi grating-coupled waveguide after oxidation and attachment of 3-APTES, Sulfo-SMCC, 16-mer probe DNA ($50 \mu\text{M}$) and complimentary PNA ($50 \mu\text{M}$). (b) Resonance shifts of all-PSi grating-coupled waveguides functionalized with 16-base probe DNA ($50 \mu\text{M}$) after exposure to complimentary PNA ($50 \mu\text{M}$), mismatch PNA ($50 \mu\text{M}$) and HEPES buffer.

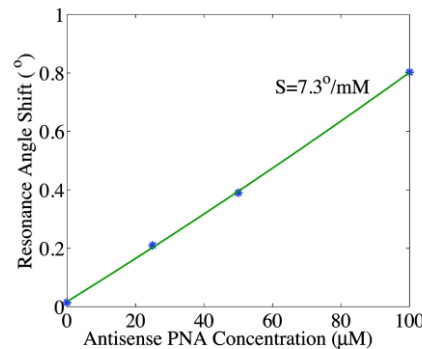


Fig. 8. Detection sensitivity estimation for all-PSi grating-coupled waveguide sensor based on resonance angle shifts due to exposure of sensor to different complementary PNA concentrations. The 16-mer probe DNA concentration for all samples represented in the figure is $100 \mu\text{M}$. The typical sample-to-sample variation in the resonance angle for each data point is $\pm 0.1^\circ$.

4.2 Photoresist grating biosensor & SOI waveguide biosensor

In order to experimentally compare the detection sensitivities of the different grating-coupled waveguide configurations considered in section 2, the same optical measurements described in section 4.1 were performed on both the photoresist grating-coupled PSi waveguide biosensor and the SOI waveguide with SiO_2 gratings biosensor. We note that although only one SOI guided mode resonance sensor was prepared for this study, the experimentally measured resonance shifts due to molecular attachment are in good agreement with our expectations based on theoretical calculations. Table 1 gives the resonance angle shifts of the three different biosensor structures after each molecule attachment step described in section 3.2. As expected, the resonance angle shifts for the photoresist grating-coupled PSi waveguide

were less than those of the all-PSi grating waveguide biosensor because the photoresist gratings inhibit molecular infiltration into the PSi regions beneath the gratings. Note, however, that the hybridization efficiency of the two types of grating-coupled PSi waveguide sensors is comparable: based on several DNA hybridization experiments, including those reported in [20], the hybridization efficiency in photoresist grating-coupled PSi waveguides is 80% - 92%. For the SOI grating-coupled waveguide biosensor that can only accommodate molecular binding on the top surface of the sensor, Table 1 shows that comparatively small resonance shifts were observed after molecular attachment, which are consistent with the theoretical calculations in section 2. The high hybridization efficiency is attributed to relatively low probe DNA coverage (~30%, according to RCWA calculations) on the surface of the SiO₂ gratings [27, 28], and ease of access of the complementary PNA molecules to the probe DNA molecules on the planar sensor surface.

Table 1. Resonance Angle Shifts after Various Molecular Attachments in All-PSi Grating-Coupled Waveguide, Photoresist Grating-Coupled PSi Waveguide, and SOI Waveguide with SiO₂ Gratings

	3-APTES	Sulfo-SMCC	Probe DNA (100 $\mu\text{mol}\cdot\text{L}^{-1}$)	Antisense PNA (100 $\mu\text{mol}\cdot\text{L}^{-1}$)
All-PSi grating	$1.70^\circ \pm 0.17^\circ$ ^a	$3.82^\circ \pm 0.07^\circ$ ^a	$0.83^\circ \pm 0.16^\circ$ ^a	0.81°
Photoresist grating	$1.47^\circ \pm 0.13^\circ$ ^b	$2.46^\circ \pm 0.44^\circ$ ^b	$0.70^\circ \pm 0.04^\circ$ ^b	$0.59^\circ \pm 0.08^\circ$ ^b
SOI-SiO ₂ grating	0.038°	0.046°	0.058°	0.062°

^aData is based on three experiments.

^bData is based on five experiments.

5. Conclusion

Due to their enhanced available surface area for molecular binding in regions of strong field localization, all-PSi grating-coupled waveguides were shown to be highly efficient biosensors. RCWA calculations demonstrated that all-PSi grating-coupled waveguides supported a 32-fold higher field distribution in regions where molecules could bind compared to planar grating-coupled SOI waveguides, and a 2-fold higher field distribution compared to photoresist grating-coupled PSi waveguides. Experiments conducted on all three types of grating-coupled waveguides similarly showed that waveguide resonance shifts due to molecular attachment were substantially larger in the all-PSi grating-coupled waveguide sensor. A small molecule detection sensitivity of 42°/RIU (1090 nm/RIU), and PNA detection limit on the order of 1 μM were demonstrated using the all-PSi grating-coupled waveguide. An improved detection limit could be achieved by increasing the probe DNA coverage using in situ probe DNA synthesis [21] or enhancing the resolution of the measurement instrument.

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