

Sub-wavelength grating for enhanced ring resonator biosensor

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Abstract: While silicon photonic resonant cavities have been widely investigated for biosensing applications, enhancing their sensitivity and detection limit continues to be an area of active research. Here, we describe how to engineer the effective refractive index and mode profile of a silicon-on-insulator (SOI) waveguide using sub-wavelength gratings (SWG) and report on its observed performance as a biosensor. We designed a 30 μm diameter SWG ring resonator and fabricated it using Ebeam lithography. Its characterization resulted in a quality factor, Q , of $7 \cdot 10^3$, bulk sensitivity $S_b = 490 \text{ nm/RIU}$, and system limit of detection $sLoD = 2 \cdot 10^{-6} \text{ RIU}$. Finally, we employ a model biological sandwich assay to demonstrate its utility for biosensing applications.

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1. Introduction

Silicon photonics is an emerging chip-scale technology focused on manipulating near-infrared light in sub-micron silicon wires [1]. Its compatibility with mainstream CMOS foundry processes facilitates the fabrication of complex chip-scale photonic systems at economies of scale [2, 3] for optical multiplexing [4], modulation [5], and biosensing [6]. While the large index contrast among the silicon waveguide, substrate, and cladding, helps to confine and guide the light, a portion of its electric field extends outside the waveguide as an evanescent field. This field is sensitive to refractive index changes outside the waveguide, such as accumulating molecules binding on its surface.

Many biosensors utilize a resonant cavity such as a ring [6–8], disk [9–11], grating [12, 13], or photonic crystal [14] for biosensing. As molecules bind to the surface, their adsorption increases the local refractive index causing a change in the mode's overall effective index resulting in a cavity's resonance wavelength shift. This change in the resonant condition can be determined using a tunable laser source and calibrated to the mass of bound molecules [15]. Silicon photonic resonant cavities have been used in a variety of clinically relevant biosensing applications including nucleic acid detection [6, 16], protein biomarkers for cancer [17, 18], viral [19, 20], and environmental toxins [21].

Ring resonators have been extensively investigated as silicon photonic biosensors due to their design simplicity and ease of fabrication. The commercially available silicon photonic biosensing platform, Genalyte, utilizes ring resonators for TE polarized light and provides a bulk sensitivity of 54 nm/RIU and detection limit of 1 ng/mL, or 10^{-5} RIU [22, 23] (for simplicity quasi - TE and - TM modes are referred to as TE and TM modes). Yet many clinical diagnostic assays require lower detection limits [24] requiring secondary amplification [25]. While achieving clinically relevant sensitivities includes both robust surface chemistries that resist fouling while allowing high densities of capture molecules and the native performance of the sensor, many groups have sought to improve the ring resonator's sensitivity.

The sensitivity is determined by the overlap of the electric field with the analyte and can be improved by increasing that overlap. Because of the high index contrast of Si/SiO₂ (or Si/H₂O) most of the electric field is confined in the core of the waveguide for TE polarized light. A way to improve the sensitivity for TE light is to decrease the waveguide thickness, as was demonstrated by Talebi Fard *et al.* with 90 nm thick waveguide that achieved a bulk sensitivity of 100 nm/RIU [26]. Both European collaborative biosensing projects SABIO and InTopSens demonstrated the use of TE polarized slot waveguides [27, 28] with a sensitivity of 212 nm/RIU and 298 nm/RIU, respectively [29, 30], and detection limits on the order of $5 \cdot 10^{-6}$ RIU. For a comparable waveguide geometry, the confinement of the TM polarized light is weaker improving sensitivities to 200 nm/RIU [22, 31]. Finally, our group demonstrated a slot waveguide Bragg grating with sensitivity of 340 nm/RIU [13].

In this paper, we report on the biosensing performance of a sub-wavelength grating (SWG) ring resonator designed for TE polarized light, resulting in a 2X sensitivity improvement over the best TM ring / slot rings. Sub-wavelength grating waveguides allow the designer to engineer the effective index of the guiding structure to minimize loss, enhance guiding capabilities, and more importantly improve the field overlap with biomolecules on the waveguide's surface. SWG gratings in silicon-on-insulator (SOI) waveguides have recently been proposed by the National Research Council of Canada (NRC) [32–34]. And while sub-wavelength gratings have been demonstrated experimentally in applications like fiber-to-chip couplers to minimize mode mismatch loss [35–38], meta material lenses [39], waveguide crossings [32], and filtering applications [40], they have yet to be experimentally demonstrated for biosensing applications [41, 42]. Wangüemert-Pérez *et al.* proposed the use of SWG waveguides for biosensing [42] and employed a Fourier-type 2D vectorial simulation tool to analyze the effect of various duty cycles on the sensing performance. They also performed a full 3D FDTD simulation to

determined the theoretical sensitivity but do not report any experimental results. Similarly, our group has also demonstrated the fabrication and measurement of ring resonators and investigated their theoretical sensing performance [43]. In this work, we show that we can further improve the performance of TE mode ring resonator and present experimental results that show our biosensors achieve 10x enhanced sensitivity over their TE-mode strip waveguide counterparts. This achievement expands their use in applications that require greater sensitivities and detection limits that what can be achieved today.

1.1. Theory of operation

Electromagnetic wave propagation in periodic media can be described by the Bloch-Floquet formalism [44–46]. Depending on wavelength, propagation can be divided into three wavelength zones for a fixed period Λ of the grating [45]: 1) the sub-wavelength zone in which the wavelength to period ratio is $\frac{\lambda}{\Lambda} > 2 \cdot n_{eff}$. This corresponds to the wavelength range longer than the Bragg wavelength and the waveguide behaves like a conventional waveguide. The periodic structure supports a true lossless mode in this case [47]; 2) The wavelength range corresponding to the photonic bandgap where Bragg reflections occur; and 3) the wavelength range shorter than the Bragg wavelength where the Bloch wave becomes leaky and part of the energy is radiated out of the waveguide and the propagation loss is determined by reflection and diffraction at the segment boundaries due to the high index contrast [48]. By having $\Lambda \ll \lambda$ the mode is without loss because the reflection and diffraction effects are suppressed. This is analogous to the electron distribution in periodic potentials, like in semiconducting materials.

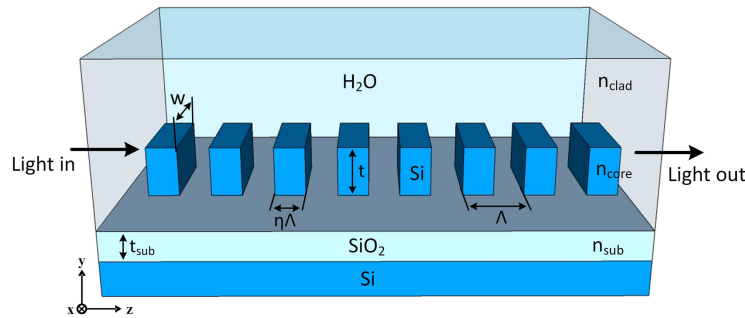


Fig. 1. Schematic of SWG waveguide: w is the waveguide width and t the thickness; Λ is the SWG period and the length of the Si blocks is determined by the duty cycle η .

For a photonic circuit designer, SWG waveguides are attractive because they allow tailored propagation properties (namely the mode shape and dispersion) by varying the duty cycle (η), period (Λ), waveguide width (w) and thickness (t). The waveguide is divided into small, slab segments (blocks of Si with refractive index n_{core}) with length $\Lambda\eta$, where Λ is the period and η is the duty cycle. The small, slab-segment cross-section is comparable to traditional waveguides. Figure 1 shows the schematic of a SWG waveguide. The substrate material with refractive index, n_{sub} , and thickness, $t_{sub} = 2 \mu\text{m}$, is SiO_2 and as cladding material, n_{clad} we used water since most biological applications require an aqueous solution.

Here we extend the theoretical analysis of SWG waveguides as biosensors and confirm the sensitivity of SWG ring resonators experimentally using a refractive index solution set. Like in the case of a uniform strip waveguide, the light is confined in the xy -plane by the index contrast (Eigenmodes). The periodicity in the z -direction ($n^2(z) = n^2(z + \Lambda)$) guarantees that the wave vector k_z is still conserved. According to the Bloch (or Floquet) theorem an electromagnetic solutions takes the form:

$$E = E_K(x, y, z)e^{-iKz} \quad (1)$$

where K is the Bloch wavenumber and $E_K(x, y, z)$ is a periodic function with period Λ so that $E_K(x, y, z) = E_K(x, y, z + \Lambda)$. Similar to the dispersion relation for regular waveguides, the dispersion relation for SWG waveguides is $\omega = \omega(K)$. The Bloch wave vector K can either be real or complex depending on the spectral regime. When K is real, the intensities of the Bloch wave will be a periodic function of position in the medium and propagate without loss. For a layered structure with uniform material properties in the xy -plane, analytical solutions exist [45, 46], but not for the case of index guided modes (vertical and lateral confinement) and numerical tools should be utilized. The effective index for Bloch modes is $n_{eff,B} = \frac{c}{\omega} K(\omega)$ and the group index is $n_g = c \frac{\partial K}{\partial \omega}$ [49, 50].

2. Simulation

For a SWG waveguide (one dimensional photonic crystal) an analytical solution does not exist and numerical methods have to be used. The most rigorous approach is a full 3D vectorial FDTD approach. However for large structures this approach is computationally very demanding and therefore not suitable for large parameter sweeps. Figure 2 shows the field magnitude along the

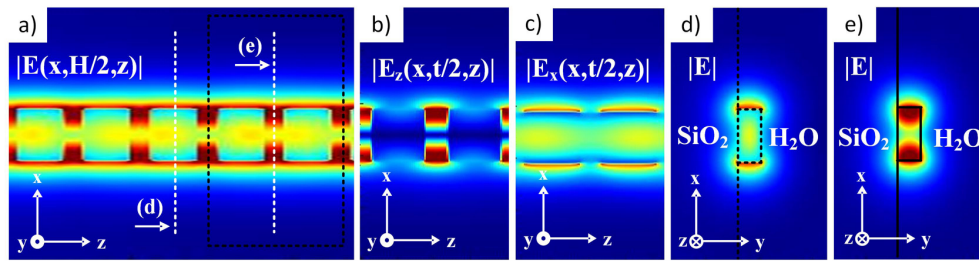


Fig. 2. a) Electric field magnitude distribution in the xz -plane defined by a cut at $y = t/2$ for a SWG waveguide with dimensions of $w = 500 \text{ nm}$, $\Lambda = 250 \text{ nm}$, $\eta = 0.7$, and $t = 220 \text{ nm}$; b) Distribution of the z -component (E_z); c) Distribution of the x -component (E_x); d) Cross-section in the middle of Si block, and e) Cross-section in the middle of the gap.

propagation axis (z -axis) for a SWG waveguide with width $w = 500 \text{ nm}$, thickness $t = 220 \text{ nm}$, period $\Lambda = 250 \text{ nm}$, and duty cycle $\eta = 0.7$. The data is reported in the xz -plane defined by a cut at $y = t/2$. Figures 2(b) and 2(c) show the magnitude of the z -component ($E_z(x, t/2, z)$) and x -component ($E_x(x, t/2, z)$) of the electrical field, respectively. Figures 2(d) and 2(e) show the mode profile in the cross section (xy -plane) of the SWG waveguide at the center of the silicon segment and in the gap respectively. The 3D simulation was performed with Lumerical's FDTD Solutions. A mode source (fundamental TE mode, $\lambda = 1.55 \text{ }\mu\text{m}$) is used to inject light into the SWG waveguide. The field is strongly confined in between the silicon segments, similar to what is observed in slot waveguides [13].

2.1. Analysis using equivalent effective refractive index

The equivalent effective index method is an efficient way to analyze the SWG waveguide. If $\Lambda < \frac{\lambda_B}{2n_{eff}}$ is true, a SWG waveguide behaves like a continuous waveguide but with a decreased refractive index. As a consequence, the light propagation can be described like in an index-guided structure. Weissman et al. showed that the "averaged" refractive index step is primarily dependent on the duty cycle and can be expressed as ([47, 51]):

$$\Delta n' = \eta \Delta n \quad (2)$$

where η is the duty cycle and Δn is the refractive index modulation of the SWG waveguide ($\Delta n = n_{core} - n_{clad}$) (similar findings by Li et al. [52] and Bierlein et al. [53]).

This reduced refractive index step results in a decreased equivalent refractive index of the waveguide and a weaker confinement of the guided mode. The refractive index of the continuous equivalent waveguide is given by: $n_{core,eq} = n_{clad} + \eta \Delta n$. This method has been used by our group and others to design the building blocks of photonic circuits such as SWG waveguides, strip to SWG mode converters, directional couplers, and ring resonators [43, 54, 55]. It needs to be noted that this approach leads to a very narrowband approximation, especially for wavelength close to the Bragg condition.

2.2. FDTD analysis

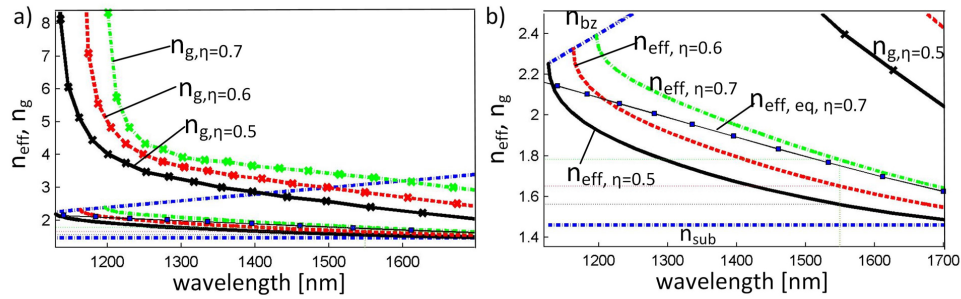


Fig. 3. a) Effective and group index of SWG waveguide with geometry: $w = 500$ nm, $\Lambda = 250$ nm, and $t = 220$ nm; The blue dash-dotted lines indicate the substrate (n_{sub}) and the Brillouin zone (n_{bz}) limits. For a duty cycle $\eta = 0.7$ the equivalent effective index approximation is shown; b) zoomed in to show the refractive index range 1.4 to 2.4 (range of effective index).

If the SWG waveguide supports multiple Bloch modes or is designed close to the photonic band gap, the equivalent model (described above) is no longer valid [33, 56, 57]. In this case, a fully vectorial 3D FDTD analysis can be done but is computationally taxing, especially for geometry sweeps. The mesh size has to be small enough to accurately resolve the periodic structure and the simulation time has to be long enough for the electromagnetic fields to fully decay. Here we make use of the periodicity, by simulating only one unit cell with Bloch boundary conditions. This approach is borrowed from band structure calculations of photonic crystals [50, 58] and we have used this method previously to determine coupling coefficients in SOI Bragg gratings [59]. Briefly, randomly distributed dipole sources are used to excite all Bloch modes in the SWG waveguide. The resonant Bloch modes are determined with a Fourier Transform of the recorded time signals for each k_z -vector (Bloch boundaries are only used in the propagation direction z). The Brillouin zone for this 1-D period structure corresponds to $-\frac{\pi}{\Lambda} < k_z < \frac{\pi}{\Lambda}$. With this method, the group index of the fundamental Bloch mode is calculated using the first order approximation as used for strip waveguides: $n_g = n_{eff,B} - \lambda \frac{\partial n_{eff,B}}{\partial \lambda}$. Figure 3 shows the effective refractive and group index for different duty cycles and a waveguide geometry $w = 500$ nm, $\Lambda = 250$ nm, and $t = 220$ nm. The effective index is limited by the photonic band gap (Bragg reflection) and the index of the substrate. At 1550 nm the effective index is $n_{eff} = 1.56$, $n_{eff} = 1.67$, and $n_{eff} = 1.71$ for duty cycles of $\eta = 0.5$, $\eta = 0.6$, and $\eta = 0.7$ respectively. The group indices for these same duty cycles were calculated to be $n_g = 2.41$, $n_g = 2.85$, and $n_g = 3.31$ respectively.

As can be seen, the group index dramatically increases as the effective index approaches the band gap of the periodic structure. As the wavelength is far away from the band gap the equivalent effective index model can be used (in Fig. 3(b) the equivalent effective index is shown for a duty cycle $\eta = 0.7$). Figure 4 depicts the effective index for different grating periods. The Bragg condition limit, $\frac{\lambda_B}{2\Lambda}$, moves to longer wavelengths as the period increases. However at the

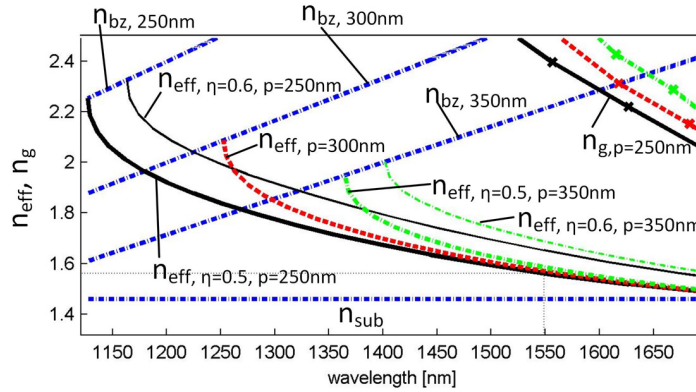


Fig. 4. Effective index of SWG waveguide with geometry $w = 500 \text{ nm}$ and $t = 220 \text{ nm}$ for grating periods $\Lambda = 250 \text{ nm}$ (black), $\Lambda = 300 \text{ nm}$ (red), and $\Lambda = 350 \text{ nm}$ (green); The blue dash-dotted lines indicate the substrate (n_{sub}) and the Brillouin zone (n_{bz}) limits. If not indicated the duty cycle of the grating is $\eta = 0.5$.

wavelength of 1550 nm the effective index is not much affected. For E-Beam lithography the minimum feature size is well below 75 nm (required for a period $\Lambda = 250 \text{ nm}$ and duty cycle $\eta = 0.7$) and SWGs can be designed far away from the bandgap. However, for foundry processes, employing deep-UV lithography, the minimum feature size is around 130 nm and designs will be much closer to the bandgap where the equivalent refractive index model is not valid anymore.

2.3. Loss

For regular strip waveguides ($w = 500 \text{ nm}$, and $h = 220 \text{ nm}$) the propagation loss for TE polarized light is around 4 dB/cm with the main loss mechanism being scattering due to sidewall roughness [60] (the SWG rings in this work are fabricated on the same direct write electron beam lithography system as used by Bojko *et al.* [60]). The propagation loss is therefore dependent on the fabrication process. The loss values reported in [60] are for strip waveguides and for obvious reasons not directly applicable for SWG waveguides. A careful loss analysis (in water as cladding material) has yet to be carried out. For SWG waveguides the optical mode is delocalized which minimizes the electric field interaction with the surface scattering sources [48], but depending on thickness of the SiO_2 layer can also increase substrate leakage. Furthermore, the sidewall interaction is reduced by $(1 - \eta)$, due to the presence of gaps between segments. But in the same time one also has to consider the roughness of the created internal sidewall. Bock *et al.* reports measured loss values of 2.5 dB/cm for a SWG waveguide with slightly narrower waveguide [48]. Furthermore, a careful loss analysis (both theoretical and experimental) should also include bending losses as function of radius.

2.4. Sensitivity analysis

Assuming the equivalent continuous waveguide model, one might expect an enhanced sensitivity compared to a strip waveguide of the same cross-section. Since the average refractive index step is reduced, the mode confinement is weaker resulting in a decreased overall equivalent effective refractive index for the guided mode. Weaker mode confinement enhances the modal overlap with the analyte (increased susceptibility), thereby increasing bulk sensitivity. However, as shown in Fig. 2, the benefit of using an SWG waveguide for sensing does not only lie in the weaker mode confinement but more in the presence of a high electric field in the gaps of the SWG structure. Therefore, a sensitivity analysis based on the equivalent index method (as

performed in [43]) needs to be verified with a proper 3D simulation method and also, maybe more importantly, verified experimentally.

The bulk sensitivity of a ring resonator can be defined as [61]:

$$S_b = \frac{\Delta\lambda_{res}}{\Delta n_{clad}} = \frac{\lambda_{res}}{n_g} \left(\frac{\partial n_{eff}}{\partial n_{clad}} \right) \quad (3)$$

The waveguide susceptibility $\frac{\partial n_{eff}}{\partial n_{clad}}$ is simulated with the equivalent effective waveguide model (using MODE Solutions) and in 3D with the Bloch boundary conditions (using FDTD Solutions). The results for duty cycle $\eta = 0.6$ and $\eta = 0.7$ are shown in Fig. 5 for a waveguide geometry of $w = 500$ nm, grating period $\Lambda = 250$ nm, and waveguide thickness $t = 220$ nm. The equivalent effective waveguide model predicts a sensitivity of around 500 nm/RIU while the 3D FDTD model with Bloch boundaries shows a sensitivity between 400 nm/RIU and 480 nm/RIU, as shown in Fig. 5.

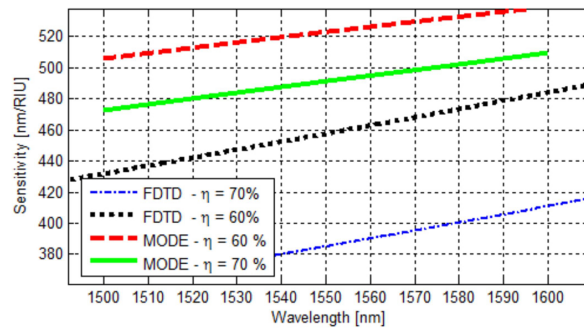


Fig. 5. Comparison of MODE Solutions and FDTD Solutions sensitivity simulations for a waveguide geometry: $w = 500$ nm, grating period $\Lambda = 250$ nm, waveguide thickness $t = 220$ nm, and duty cycle of $\eta = 0.6$ and $\eta = 0.7$ respectively.

In a similar way one can define the surface sensitivity as:

$$S_s = \frac{\Delta\lambda_{res}}{\Delta t_{ad}} = \frac{\lambda_{res}}{n_g} \left(\frac{\partial n_{eff}}{\partial t_{ad}} \right) \quad (4)$$

where t_{ad} is the uniform thickness of the adsorbed protein layer with refractive index n_{ad} . For most proteins the refractive index is around $n_{ad} = 1.48$ [62, 63]. The thickness of the layer is determined by the characteristic length of the protein and the orientation of the adsorbed protein. For globular shaped proteins the characteristic length corresponds to the diameter of a sphere. The refractive index of this layer is also dependent on the surface density of the protein. For a surface with all possible sites occupied by a protein, the refractive index will be $n_{ad} = 1.48$. In reality, not all surfaces are fully covered and so one can assume a constant refractive index of the layer, $n_{ad} = 1.48$, with a changing effective thickness $t_{eff} < t_{ad}$ since the thickness changes linearly with surface concentration. This approach has been proposed and used in [64, 65]. The 3D FDTD simulation with Bloch boundary conditions estimates the susceptibility $\frac{\partial n_{eff}}{\partial t_{ad}}$ to around $1.45 \mu\text{m}^{-1}$ ($w = 500$ nm, grating period $\Lambda = 250$ nm, waveguide thickness $t = 220$ nm, and duty cycle of $\eta = 0.6$) which is about 3X compared to a regular quasi TE waveguide [22].

3. Experimental approach

3.1. Design and fabrication

SWG photonic circuits were realized on a 220 nm thick SOI wafer (SOITec, Grenoble, France) using a JEOL JBX-6300FS Direct Write E-Beam Lithography System (EBeam) at the University

of Washington's Nanofabrication Facility (WNF). The EBeam provides a low-cost, fast turn-around fabrication process that has been optimized to produce consistent, robust, passive silicon photonic components [60, 66]. A SWG ring resonator sub-circuit consists of slab-to-SWG converter [54], straight-to-bent SWG waveguide directional coupler, looped-back bent SWG waveguide to create the resonant structure. The coupling coefficients for the directional coupler were estimated from [54] and multiplied by a factor 2, to take into account the H₂O cladding instead of air (factor 2 based on strip waveguide simulations). SWG rings with radius of 20 μm and 30 μm , duty cycles $\eta = [0.5, 0.6, 0.7]$, and coupling gaps $g = [200, 250, 300, 350, 400]$ nm were fabricated and characterized to determine the optimal design parameters for biosensing applications. Figure 6 shows a SEM image of a SWG ring resonator with the following design parameters: waveguide width $w = 500$ nm, grating period $\Lambda = 250$ nm, waveguide thickness $t = 220$ nm, and duty cycle $\eta = 0.7$.

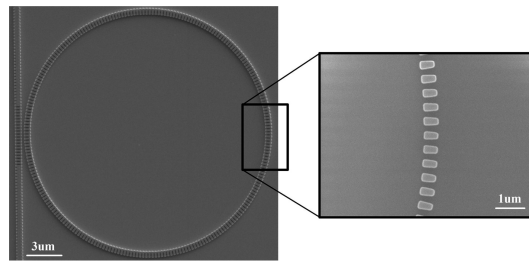


Fig. 6. SEM image of SWG ring resonator fabricated by Ebeam lithography. Waveguide geometry: $w = 500$ nm, grating period $\Lambda = 250$ nm, waveguide thickness $t = 220$ nm, and duty cycle $\eta = 0.7$.

3.2. Microfluidics and assay control

Devices were tested and characterized using a custom test setup and software application [55]. Briefly, a custom fiber array from PLC Connections (Columbus, OH) with four polarization maintaining fibers is used to couple the light on and off the chip through on-chip vertical gratings [38]. An Agilent 81682A 1.5 μm tunable in a 8164A mainframe together with Agilent's N7744A power meters is used as off chip laser source and detection equipment. The chip under test sits on an a thermally tuned aluminum chuck. The temperature is controlled with a Stanford Research LDC501 controller (Stanford, CA). Two motorized stages (Micos, Germany) are used to align the chip to the fiber array. To sequence reagents from a 96-well plate additional two motorized stages are used (Velmex XSlide, Bloomfield, NY). A silicon gasket with defined 500 μm wide channels is secured by a PTFE flow cell. The silicon gasket has a thickness of 500 μm . A syringe pump (Chemiyx Nexus 3000, Houston, Tx) is used to control flow rates by drawing reagents from the wells over the sensors. Finally, a custom application written in MATLAB provides instrument control, orchestrates the acquisition sequence, and processes each acquired data set for real-time resonant peak tracking during the assay.

3.3. Reagents

The SWG ring resonator's performance, including quality factor Q , intrinsic limit of detection, $iLoD$, and bulk sensitivity S_b , were characterized in an aqueous environment using NaCl refractive index standards ranging from 62.5 mM to 1 M. The RI of each solution was measured using a Reichert AR200 digital refractometer (Depew, NY). A standard sandwich assay involving well characterized reagents were used to evaluate the sensor's utility for biosensing applications [55]. Protein-A, a 42kDa membranous protein observed to preferentially bind an

immunoglobulin's Fc domain, was obtained from ThermoFisher (Chicago, IL). The immunoglobulin, anti-Streptavidin (antiSA) (MW 150kDa), and its conjugate ligand, Streptavidin (SA) (MW 57kDa), were obtained from Vector Labs (Burlingame, CA). For amplification, biotin was conjugated to BSA (MW 66kDa) using a kit from Bangs Labs (San Diego, CA). A FisherScientific 1x PBS solution (Hampton, NH) was used to dilute reagents and rinse unbound molecules after each step.

4. Results and discussion

4.1. Optical spectrum

Figure 7 shows the measured transmission spectrum of a SWG ring with radius $R = 30 \text{ } \mu\text{m}$, waveguide width $w = 500 \text{ nm}$, thickness $t = 220 \text{ nm}$, SWG period $\Lambda = 250 \text{ nm}$, and duty cycle of $\eta = 70 \%$ for two different coupling gaps, 300 nm and 400 nm respectively. The linewidth

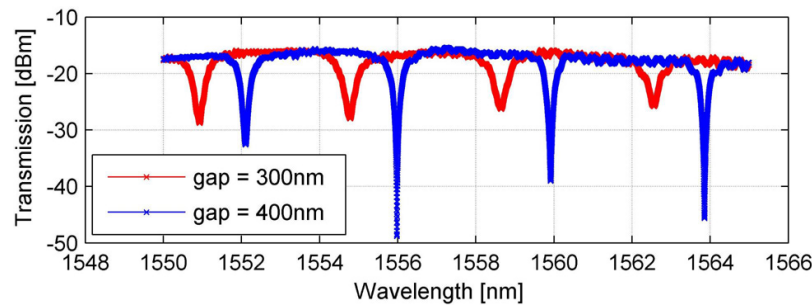


Fig. 7. a) Transmission spectrum for a ring with $R = 30 \text{ } \mu\text{m}$, $w = 500 \text{ nm}$, $t = 220 \text{ nm}$, $\Lambda = 250 \text{ nm}$, $\eta = 70 \%$, and gaps of $g = 300 \text{ nm}$ and $g = 400 \text{ nm}$ respectively, exposed to DI water.

of the resonator with a gap of 400 nm is about 0.2 nm which corresponds to a quality factor Q of about $7 \cdot 10^3$. The linewidth is defined as the full width at half-maximum, FWHM. The group index, n_g , of the waveguide can be extracted from the free spectral range (FSR) of a ring resonator given by $n_g = \frac{\lambda^2}{FSR \cdot L}$ with L the ring resonator round trip length ($L = 2R\pi$). For the SWG ring resonator with radius $R = 30 \text{ } \mu\text{m}$ and duty cycle $\eta = 0.7$, the $FSR = 3.936 \text{ nm}$ and the corresponding group index $n_g = 3.27$. For a ring with the same radius but different duty cycle of $\eta = 0.6$, results in a $FSR = 4.54$ and group index of $n_g = 2.81$. These observed results agree with the simulated (FDTD) values for $n_g = 3.31$ and $n_g = 2.85$, respectively.

4.2. Bulk RI sensing

Figure 8(a) shows the measured transmission spectrum of the SWG ring exposed to ultrapure water and the refractive index solution standards. The chip stage is thermally tuned to 25°C to limit thermal drift. For each concentration the optical spectrum was measured ten times to ensure repeatability and signal stability. The resulting stepped, resonant wavelength shifts are shown in Fig. 8(b). The slope of the resonant wavelength shifts per refractive index standard is the bulk sensitivity of the sensor, as shown in Fig. 9. The SWG ring with duty cycle $\eta = 0.6$ yields a bulk sensitivity of $S_b = 491 \text{ nm/RIU}$ (see Fig. 9) and the sensor with a duty cycle of $\eta = 0.7$ yields $S_b = 405 \text{ nm/RIU}$. Both of these values are reported for a wavelength of around $\lambda = 1575 \text{ nm}$. Compared to regular TE ring resonators [6, 18] SWG ring resonators show an 8X increase in sensitivity, and compared to TM [55] a 2X increase. SWG ring resonators provide close to twice the bulk sensitivity as slot resonators (Bragg and ring [13]). While the equivalent waveguide

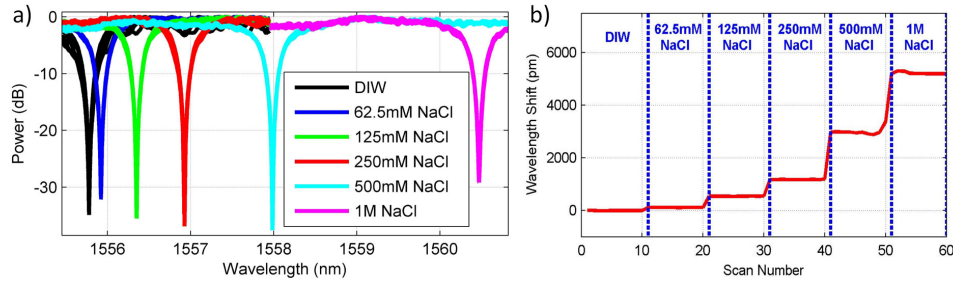


Fig. 8. a) Transmission peak for a ring with $R = 30 \mu\text{m}$, $w = 500 \text{ nm}$, $\Lambda = 250 \text{ nm}$, $\eta = 70 \%$, $t = 220 \text{ nm}$, and gap $g = 400 \text{ nm}$ exposed to different solutions of NaCl; b) reported peak wavelength shift.

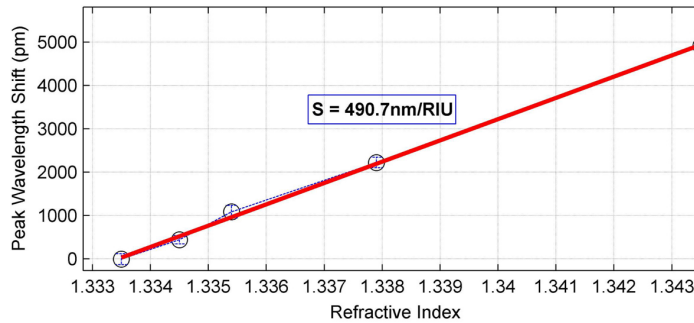


Fig. 9. Sensitivity results for a SWG ring with $R = 30 \mu\text{m}$, $w = 500 \text{ nm}$, $\Lambda = 250 \text{ nm}$, $\eta = 60 \%$, and $t = 220 \text{ nm}$

model tends to overestimate the bulk sensitivity (the simulated sensitivities are shown in Fig. 5, the measured sensitivities compare well to the simulated results).

The measured Q factor is $7 \cdot 10^3$, and sensitivity is $S_b = 405 \text{ nm/RIU}$, the intrinsic limit of detection is $iLoD = \Delta n_{min} = \frac{\lambda}{Q S_b} = 5.5 \cdot 10^{-4} \text{ RIU}$. The intrinsic limit of detection has been proposed by Chrostowski *et al.* [61] as a figure of merit independent on read out circuitry and data processing. The theoretical limit for the intrinsic limit of detection for an ideal resonator sensor (in water) operating at $\lambda = 1.55 \mu\text{m}$ is $2.4 \cdot 10^{-4} \text{ RIU}$ [61]. The measured minimal detectable wavelength shift of the SWG resonator (and system) presented here is $\Delta\lambda_{min} = 1 \text{ pm}$, which translates into a system detection limit $sLoD = \frac{\Delta\lambda_{min}}{S_b} = 2.47 \cdot 10^{-6} \text{ RIU}$ for the resonator with a duty cycle $\eta = 0.7$ and $sLoD = 2 \cdot 10^{-6} \text{ RIU}$ for the resonator with a duty cycle $\eta = 0.6$.

4.3. Biosensing

A sandwich assay involving well characterized biomolecules with high binding affinities were used to evaluate the biosensing performance of the SWG ring. Figure 10 depicts the sequenced steps of the assay which include: (1) physisorbing an initial adlayer onto the waveguide surface, (2) immobilizing antibodies to the adlayer, (3) blocking any remaining exposed surfaces on the sensor to limit 'biological noise' resulting from subsequent, non-specific interactions, (4) capturing the target antigen, and (5) using a secondary molecule (label) to amplify the captured target.

The experimental results in Fig. 11 show the resonant wavelength shifts over the course of the assay and demonstrate the sensor's ability to detect molecular binding events in real time.

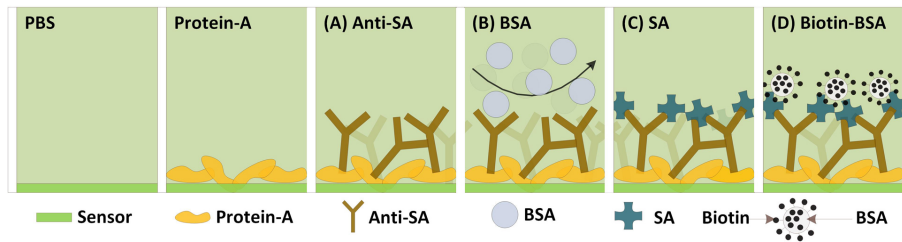


Fig. 10. Biosensing cartoon: Region A = anti-streptavidin (antiSA) (10 $\mu\text{g/mL}$), B = Bovine Serum Albumin (BAS) (20 $\mu\text{g/mL}$), C = streptavidin (SA) (20 $\mu\text{g/mL}$), and D = biotinylated-BSA (50 $\mu\text{g/mL}$). After each reagent the sensor is washed with PBS, indicated by the grey area mid-way through each region.

Protein-A (1 mg/mL) was first passively adsorbed to the sensor's native oxide surface by soaking the chip overnight. Protein-A, a 42kDa globular protein has a reported diameter of around 3 nm and refractive index of $n = 1.48$ [63]. Coen *et al.* observed that the initial adsorbed Protein A may denature resulting in a 1 nm thick add - layer but that subsequent adsorption of protein A forms a bioactive layer which provides the necessary binding pocket for the antibody's Fc domain [62, 67]. For the second and subsequent layer, we presume that the protein retains its 3 nm globular shape due to the observed bio activity, resulting in a 3-4 nm thick cladding layer ($n = 1.48$). We have shown in previous publications that our experimental and simulated results are within the range observed and reported by Coen *et al.* [9, 55]. After achieving a baseline in PBS buffer at 37°C, the biomolecules (as depicted in Fig. 10 step (A) through (D)) were sequenced over the sensor. Each reagent was followed by a 20 minute PBS buffer rinse to remove any unbound species on the sensor or in the channel (as depicted by the grey area labeled 'PBS').

The capture antibody, anti-Streptavidin (antiSA, 10 $\mu\text{g/mL}$), was introduced to functionalize the sensor for Streptavidin detection. As can be seen in Fig. 11, Region A, the adsorption of this 160 kDa protein results in a 1.5 nm resonant wavelength shift. The linear increase suggests that the affinity reaction is transport limited rather than reaction limited. While a fully saturated surface is desired, the automated test setup used to sequence reagents and monitor the wavelength shifts did not permit extending the adsorption time during the assay. Regardless, the post-acquisition simulations and experimental results observed in Fig. 11 suggest ample surface coverage for subsequent binding events.

To show that subsequent molecular binding interactions are specific, BSA (20 $\mu\text{g/mL}$) was introduced to block any remaining exposed sensor area, as shown in Region B. The slight wavelength shift offset at the end of the Region B rinse step suggests adsorption of BSA as a result of incomplete coverage of the Protein A - antiSA complex [62].

Next, the sensor is subjected to 10 $\mu\text{g/mL}$ of streptavidin (SA) as shown in Region C. The permanent resonant shift observed during the PBS rinse cycle suggests specific and irreversible binding of SA to antiSA. This step also indicates that the antiSA bound to the Protein A retains its bioactivity and specificity. The final step (Region D) further demonstrates the biological specificity of the captured SA while amplifying overall wavelength shift.

4.4. Validation model

Although the relative resonance wavelength shifts for each new add layer correspond to refractive index changes based on the respective molecular weights of each biomolecule [62, 63, 67], the results from the FDTD simulations with varying adlayer thicknesses with refractive index $n_{ad} = 1.48$ (to model protein layers) were used to provide insight into the amount of adsorbed protein for the initial adlayers. From simulation we predict a surface sensitivity of 800 pm/nm (duty cycle of $\eta = 0.6$, $n_g = 2.81$, and $\lambda = 1550$ nm). This means that the resonant wavelength

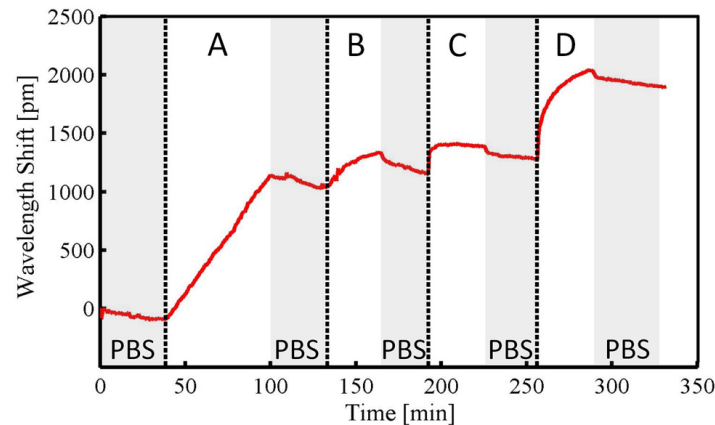


Fig. 11. Biosensing experimental results: Region A = anti-streptavidin (antiSA) (10 $\mu\text{g/mL}$), B = Bovine Serum Albumin (BAS) (20 $\mu\text{g/mL}$), C = streptavidin (SA) (20 $\mu\text{g/mL}$), and D = biotinylated-BSA (50 $\mu\text{g/mL}$). After each reagent the sensor is washed with PBS, indicated by the grey area mid-way through each region.

shifts 800 pm for each nm of adsorbed protein layer onto the sensor's surface. Since for protein adlayers the surface coverage (SC) is usually not 100% we assume a constant refractive index of the layer, $n_{ad} = 1.48$, but a changing effective thickness $d_{ad}(SC)$. We further also use the assumption that this variation is linear [63]. Assuming an antibody can be approximately modeled as a 5 nm diameter cylinder that is 10 nm tall, a shift of 1 nm (Region A) suggests a 1.25 nm uniform adlayer, or approximately 12.5% surface coverage considering the antibody model. This result agrees well with previously observed surface coverage values for antibodies [9, 62]. A shift of 500 pm (Region C) suggests a uniform adlayer of 0.6 nm for SA. Assuming SA's mass is approximately 1/3 of antiSA and 5 nm in diameter, the observed results suggest 13% surface coverage, which agrees well with the observed antiSA coverage. The 700 pm shift in Region D suggests a uniform adlayer of 0.87 nm on the sensor's surface. Even though SA is tetravalent, it is unlikely all four binding sites are occupied with bBSA. The observed wavelength shift (and resulting adlayer thicknesses) ratio of 1.4:1 for bBSA:SA is consistent with this hypothesis. Passivating the surface with BSA (Region B) provides confidence that these binding interactions are specific and not precluded by sterics. In addition to validating the bioactivity and specificity of the captured species, this sandwich assay illustrates the utility of SWG rings for multi-layer biological assays.

5. Conclusion

This paper demonstrates the first experimental demonstration and the use of SWG rings for biosensing, resulting in a 2X sensitivity increase over our best TM ring and slot waveguide resonators. The reported sensitivities of 400 - 500 nm/RIU expand the use of these devices in sensing applications requiring sensitivity and detection limits beyond today's silicon photonic ring resonator systems. And while this paper describes TE polarized light SWG rings, future work will focus on extending the simulation methods discussed here to include TM polarized resonators as well. Furthermore, to commercialize a biosensor using today's CMOS foundries, fabrication processes must achieve the minimum feature sizes required to fabricate these structures (75 nm). While these resolutions are not achievable today, these SWG ring resonators could be designed closer to their Bragg condition (eg: with a larger period, Λ) at expense of increased loss and larger group indexes.

Acknowledgments

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