

Silicon-on-insulator sensors using integrated resonance-enhanced defect-mediated photodetectors

Sahba Talebi Fard,^{1,*} Kyle Murray,¹ Michael Caverley,¹
Valentina Donzella,¹ Jonas Flueckiger,¹ Samantha M. Grist,¹
Edgar Huante-Ceron,² Shon A. Schmidt,³ Ezra Kwok,⁴
Nicolas A. F. Jaeger,¹ Andrew P. Knights,² and Lukas Chrostowski¹

¹ Department of Electrical and Computer Engineering, The University of British Columbia
2332 Main Mall, Vancouver, BC, V6T 1Z4 Canada

² Department of Engineering Physics, McMaster University
1280 Main Street West, Hamilton, ON, L8S 4L7 Canada

³ Department of Bioengineering, The University of Washington
1705 NE Pacific Street, Seattle, WA 98195 USA

⁴ Department of Chemical and Biological Engineering, The University of British Columbia
2360 East Mall, Vancouver, BC, V6T 1Z3 Canada

*sahbat@ece.ubc.ca

Abstract: A resonance-enhanced, defect-mediated, ring resonator photodetector has been implemented as a single unit biosensor on a silicon-on-insulator platform, providing a cost effective means of integrating ring resonator sensors with photodetectors for lab-on-chip applications. This method overcomes the challenge of integrating hybrid photodetectors on the chip. The demonstrated responsivity of the photodetector-sensor was 90 mA/W. Devices were characterized using refractive index modified solutions and showed sensitivities of 30 nm/RIU.

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1. Introduction

Realizing a complete lab-on-chip has attracted increasing attention due to its advantages of being compact, inexpensive, and disposable. Silicon photonic sensors on the silicon-on-insulator (SOI) platform provide a viable path towards achieving this. Recent demonstrations of silicon photonic based sensors have shown great potential for a variety of sensing applications. These include, but are not limited to, oil and gas sensing, water and air quality monitoring, and label-free detection of bio-molecules [1–8]. However, the commercial viability of these devices has been stymied, in part, due to difficulties in integrating these devices with detectors.

Silicon photonic sensors, which use near-infrared light confined to nanometer-scale silicon waveguides, exploit the interaction between the evanescent field of the waveguide mode and the material surrounding the waveguide core (e.g. analyte in the cladding medium) for detection. A change in the refractive index (RI) of the material in the cladding changes the effective index of the waveguide mode. This effect can be used to detect changes in the optical properties of an analyte under study by introducing it to the cladding medium of a waveguide. When waveguides are used in a resonant structure, a small change in the external RI causes the resonance frequencies of the resonator to shift, providing a measurable response.

Various types of silicon photonic resonators have been previously demonstrated with sensitivities sufficient for bio-molecule detection [1, 2, 9–11], however, these demonstrations did not use an integrated photodetector. Hybrid photodetectors have been demonstrated using III-V materials [12–14] or germanium [15–19] on the SOI platform. However, these detectors required the use of relatively complicated and expensive processes, which is not in keeping with the goal of providing an inexpensive lab-on-chip. To avoid the need for implementing such hybrid photodetectors, we suggest the use of defect-mediated based photodetectors on SOI [20–24]. In this paper, we propose integrating such photodetectors [20–24] with well-developed and characterized resonator-based sensors [1, 2, 9–11]. Here, we investigate and demonstrate a ring resonator sensor with an integrated detector, which exploits the resonance enhancement of the ring for increased responsivity.

2. Resonator sensors and defect-mediated photodetectors

In this section, we review the performance and characterization of defect-mediated photodetectors as well as resonator sensors. Further, we discuss the novel integration of these photodetectors with a resonator-based sensor.

2.1. Defect-mediated photodetectors

Silicon is a relatively poor material for photodetection in the near-infrared because the photon energy is less than the silicon bandgap resulting in negligible optical absorption. To overcome this, defect-mediated silicon waveguide detectors have been proposed and characterized [20–25]. These photodetectors are formed by creating a *p-i-n* diode across a waveguide

structure. An inert ion implantation into the intrinsic region causes damage to the silicon lattice which introduces deep levels in the bandgap through which light with energy less than that of the bandgap can be absorbed. The lattice damage-induced absorption, which increases the propagation loss to 20-100 dB/cm [22], causes the absorbed light to produce electron hole pairs that can be extracted by applying a reverse bias to the *p-i-n* diode. A subsequent low temperature anneal can be used to partially repair the damage to achieve an optimum defect concentration. The increased absorption due to the presence of the defects is modest, so, if a linear detector were to be used, it would have to be long in order to absorb a large fraction of the incident light and, thereby, achieve a high responsivity. Alternatively, the effective responsivity can be increased by using a resonant structure which makes use of the intensity buildup within the resonator to achieve increased absorption within a small footprint [26]. In these resonant detectors, the increased optical loss associated with the defects reduces the quality factor and therefore introduces a tradeoff between the responsivity and the precision of sensing. Defect-mediated silicon photodetectors fabricated in this manner have shown responsivities of 0.5 to 10 A/W [25].

2.2. *Evanescent field-based resonator sensors*

Evanescent Field (EF) sensors operate by detecting a change in the effective refractive index of a waveguide [6–8, 27] caused by changes in the concentration of an analyte in the cladding (bulk sensing) or as molecules bind to a functionalized surface of the waveguide core (surface sensing). Optical resonators, having small footprints and longer interactions with the analytes in their cladding media (resulting in higher sensitivities), are promising candidates for sensing applications. SOI-based resonator sensors with small footprints, each integrated with a detector, would make high-throughput and/or simultaneous analysis of multiple analytes possible, where many sensors, each designed to measure a different target could be integrated on a single chip. These chips could be designed, with various sensors aligned, to use microfluidic channels to deliver various targets to the sensors, all in a single system. Various resonator sensors have been developed and characterized, some of which include disk resonators [10]; strip waveguide and slot waveguide ring resonators [2]; strip waveguide Bragg gratings [1]; and slot waveguide Bragg gratings [9]. These resonator sensors were shown to have sensitivities and limits of detection approaching the theoretical maximum value (based on the case of a resonator in water). We have also characterized and validated these sensors using a standard ‘sandwich’ assay [1]. Having fabricated these sensors using a CMOS compatible SOI wafers, through a Multi-Project Wafer (MPW) foundry, provides the potential for integration of these sensors with on-chip electronics, and makes the integration of a complete system on a chip possible.

2.3. *Defect-mediated photodetector resonator sensor*

We propose integrating defect-mediated photodetectors with previously demonstrated resonator-based biosensors, such as ring [11], disk [10], and phase-shifted Bragg grating resonators [9], to create a biosensor with integrated photodetectors (photodetector-sensor). Figure 1 shows schematic drawings of how defect mediated photodetectors could be integrated with each type of biosensor. A top view of each sensor shows the doped and detection regions. In each case the defect-mediated detector is introduced across the waveguide that forms the resonator cavity.

The 3D rendering below each schematic view provides perspective on how the defect region and detector can be oriented in a fluidic channel for biosensing applications. The biomolecules would be introduced across the entire device area but only those in proximity to the resonator waveguide are interrogated. The ring resonator, Fig. 1(a), and disk resonator, Fig. 1(b), share similar configurations except that the disks are not etched in the center and only have one

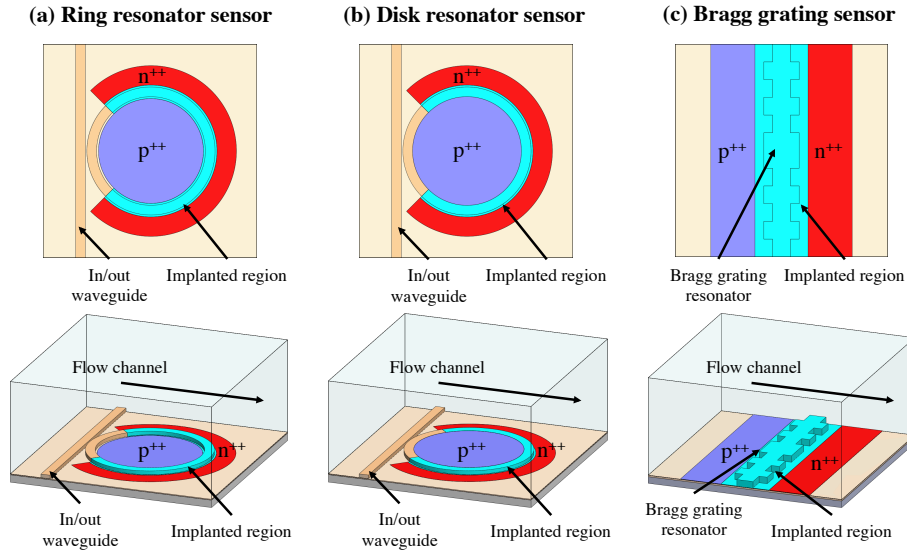


Fig. 1. Schematic representations of proposed designs for biosensing: a) a defect-mediated ring resonator photodetector-sensor; b) a defect-mediated disk resonator photodetector-sensor; and c) a defect-mediated Bragg grating photodetector-sensor. The region of the detector with defects is highlighted in bold cyan. The remaining waveguide is shown in beige.

exterior side-wall. The ring, however, has two sidewalls formed at the inner and outer radius of the ring. The n^{++} implants are shown in red while the p^{++} implants are shown in blue. This forms a PIN structure, with the detection occurring in the “I” region which has defects.

The operation of each type of defect-mediated photodetector resonator sensor is very similar. The defect-mediated photodetector in the resonance cavity senses the optical power. Since the buildup of power inside the resonator is most significant at the resonance wavelength, by measuring the photocurrent as the wavelength is varied, one can find the wavelength for which the intensity buildup is maximum and determine the resonance wavelength. Furthermore, since each resonator forms an evanescent field-based sensor as described in Section 2.2 the resonance wavelength shifts as the refractive index of the cladding changes. Thus, in this configuration the resonator allows the device to sense refractive index changes while the integrated detector allows the simultaneous readout of the sensor response in one device.

3. Design considerations and layout

As a proof of concept, two variations of ring resonator photodetector sensors like that depicted in Fig. 1(a) were designed and fabricated. A cross-section of the detector is shown in Fig. 2. In order to integrate the photodetector with the ring resonator, rib waveguides, as opposed to strip waveguides, are used to allow for electrical contact with the junction to extract photocarriers that are generated in the waveguide. Metal contacts are made to the highly doped n^{++} and p^{++} slab regions far away from the optical mode that is confined to the waveguide such that the metal does not induce significant loss. The cladding oxide is removed above the resonator so that the analyte test solution can interact with the optical mode.

We fabricated two device variations (designs A and B) that have different tradeoffs between sensitivity, responsivity, and quality factor. The two resonator designs are shown schematically

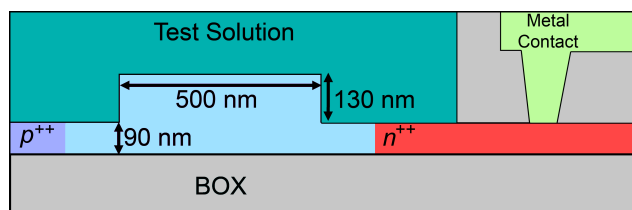


Fig. 2. Schematic of detector cross-section

along with microscopic images of fabricated devices in Fig. 3.

In design A, the oxide cladding is removed over 75% of the ring area to expose the ring to the analyte. Electrical contact from the metal routing to the ring's interior is made through a contact via in the remaining oxide cladding as shown in Fig. 3(a). This design allows contact to be made to the center of the ring without introducing significant optical loss because the metal trace only crosses over the waveguides when it is above the cladding oxide, at the expense of a reduced sensitivity due to the smaller region over which the analyte interacts with the optical mode. In design B, the oxide cladding is removed over the entire ring, so the analyte can interact with the entire ring. In this case electrical contact to the silicon is made through a via outside the ring and contact is made to the ring interior by a conductive trace formed of highly doped silicon shown as the region of p^{++} doping crossing through the ring waveguide in Fig. 3(c). In this design a greater sensitivity can be achieved by allowing the optical mode to interact with the analyte over the entire length of the ring. This benefit comes at the cost of an increase of absorption loss and lower quality factor because the highly doped trace must cross the waveguide.

In both designs A and B the resonator is in a racetrack configuration with radius $40\text{ }\mu\text{m}$ and coupling lengths are $4.2\text{ }\mu\text{m}$ and $5.1\text{ }\mu\text{m}$, respectively, each with a 200 nm coupling gap. The racetracks were designed for critical coupling using Lumerical MODE Solutions and analytical models in MATLAB. The bend and mode-mismatch losses were simulated using MODE, and propagation loss was assumed to be around 3 dB/cm [28]. The induced loss due to implanted defects and proximity of highly doped regions were estimated to be around 10 dB/cm to 30 dB/cm [29, 30]. A few coupling lengths were calculated based on these assumed values of losses (where the assumed loss due to defects were varied from 10 to 30 dB/cm). For design A, a coupling length of $4.2\text{ }\mu\text{m}$, corresponding to the assumed average loss of 20 dB/cm due to defects, resulted in a spectrum close to critical coupling. In this design, 75% of the resonator is implanted. For design B, a coupling length of $5.1\text{ }\mu\text{m}$, corresponding to the assumed loss of 30 dB/cm , resulted in a spectrum close to critical coupling. In this design, the entire ring is implanted, and in addition a trace of p^{++} doping crosses the waveguide and therefore higher losses are expected compared to design A. The bus and resonator waveguides are 500 nm wide rib waveguides, formed by a 130 nm thick waveguide on a 90 nm thick slab. As shown in Figs. 2 and 3, the p^{++} and n^{++} doping regions are 500 nm from the edges of the waveguides. We choose the distance from the waveguide to the p^{++} and n^{++} contact regions based on two considerations: the doped regions must be sufficiently close to the waveguide to extract the generated carriers efficiently before they undergo carrier recombination, and the distance must be large enough so that the optical loss from the interaction with the highly doped regions is not too high.

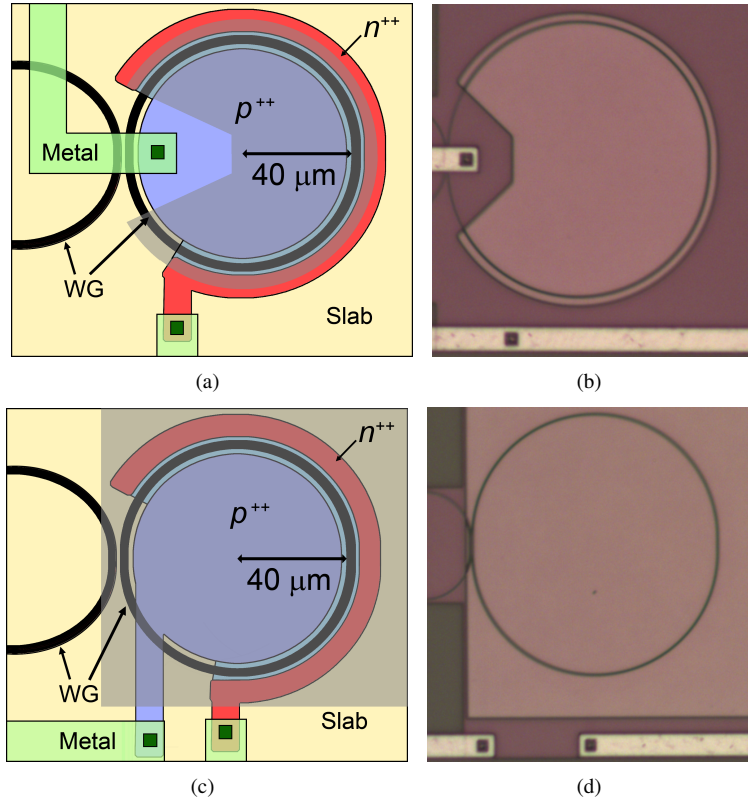


Fig. 3. a) Schematic representation of design A (where the metal contact passes above the coupling region of the ring) ; b) Microscopic picture of design A; c) Schematic representation of design B (where the highly-doped silicon wire passes through the waveguide); d) Microscopic picture of design B. In Figs (a) and (c), the gray overlay indicates where the cladding oxide was removed.

4. Experimental methods and materials

The devices were fabricated at The Institute of Microelectronics in Singapore, up to the point at which the ion implantation was performed to introduce defects in the waveguide. For this step, implantation was performed at McMaster University. Boron ions were implanted at an energy of 30 keV and dose of $5 \times 10^{12} \text{ cm}^{-2}$. No post-implantation high temperature anneal was performed such that the boron remained inactive and thus only structural defects were introduced to the waveguide (i.e. no chemical doping took place). On-chip grating couplers (GCs) facilitate the efficient coupling of light into and out of the SOI chip [31], and electrical contact to the detectors is made by routing metal traces to the device from large on-chip contact pads that can be externally probed. This sensor was characterized using the techniques and reagents described in this section.

4.1. Experimental setup

Our measurement setup used: a tunable laser (Agilent 81682A, Agilent Technologies, Inc., USA) with a wavelength range of 1460 nm to 1580 nm as the optical source; an optical power sensor (Agilent 81635A, Agilent Technologies, Inc., USA) to measure the output light intensity;

and a source measurement unit (Keithley 2602, Keithley Instruments, USA) to perform the electrical characterization. Light was coupled into and out of the chip using GCs and an array of polarization maintaining (PM) optical fibers (PLC Connections, LLC., USA). The PM fiber array was aligned to the GCs on the chip using a motorized stage that was controlled using an automatic alignment routine, to obtain maximum optical coupling to the devices. Electrical contact to the detectors was made by routing metal traces to the device from large on-chip contact pads that were externally probed, using a needle.

4.2. Reagents and microfluidic setup

To determine the bulk sensitivity, and to characterize the performance of our photodetector-sensors, standard aqueous solutions of sodium chloride (with concentrations of 0, 500 mM, 1 M, and 2 M) were characterized using a Reichert AR200 Digital Refractometer (Depew, NY). Polydimethylsiloxane (PDMS) microfluidic channels, with widths and heights of 200 μm and 80 μm , respectively, were aligned to the sensors and reversibly bonded on the chip to facilitate the delivery of standard solutions to the cladding media of our photodetector-sensors. Prior to bonding the PDMS to the chip, inlet and outlet holes for microfluidic channels were punched in the PDMS using a 0.5 mm coring tool (Schmidt Press, CORP, USA). After bonding the PDMS to the chip, the channels were linked to tygon micro-bore tubing using 21 gauge blunt needles (Nordson EFD, CORP., USA) through the punched inlet and outlet holes. These tubes were used as the fluidic inlet and outlets and were connected to a syringe. To maintain a constant flow rate of the reagents over our photodetector-sensor, a syringe pump (KDS-230, KD Scientific, Inc., USA) was set to operate in withdraw mode (negative pressure) at a constant rate of 19 $\mu\text{L}/\text{min}$.

5. Performance analysis and results

Prior to device characterization, the detectors were forward biased to produce a forward current of 1 mA for 5 minutes [32]. The current-voltage measurement for both designs is plotted in Fig. 4. The implanted photodetector-sensors were first tested under various reverse bias voltage (-2 to -35 V) conditions to determine a suitable operation point. In order to achieve a reasonable signal-to-noise (SNR), without unduly stressing the device, we chose a bias voltage of -10 V for operation.

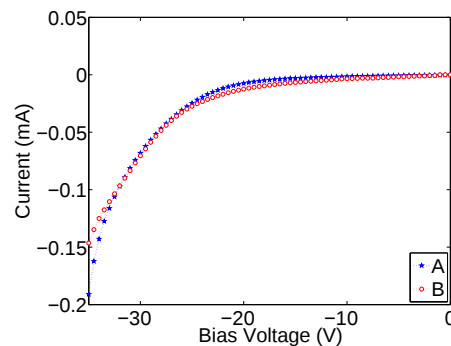


Fig. 4. IV curves for designs A and B.

To further characterize the devices, we measured both the optical and the electrical responses of our photodetector-sensors; to do this we swept the laser's wavelength while simultaneously measuring the optical power transmitted through the ring and the current generated by the

integrated detector (with a bias of -10 V). The optical power at the output of the laser was 0 dBm. There are various losses that contribute to the overall loss in the system. The primary dominant loss is associated with coupling from the fibers to the waveguides through the grating couplers. There are long (> 5 mm) routing waveguides from the input GC to the device and from device to the output GC which also contribute to the loss. However, due to the symmetry of the design, the loss from the output of the optical power to the device is equal to the loss from device to the detector connected to the output fiber. Therefore, the total loss is divided by two to approximate the power level of the optical signal that reaches the device. Figure 5 shows the measured electrical and optical response of design A and design B devices. It can be seen that when the ring is on resonance there is an increase in the photocurrent due to the resonant buildup of energy in the ring.

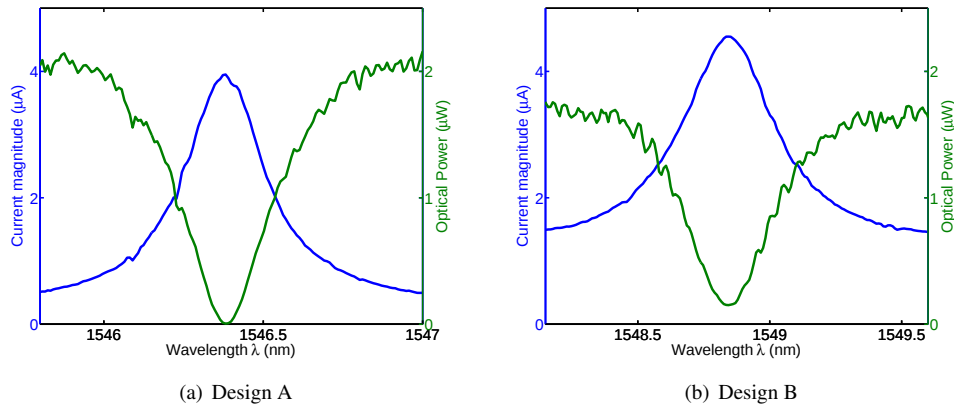


Fig. 5. Measured electrical and optical responses of our implanted detector resonator sensors: a) Design A, and b) Design B.

To estimate the responsivity of a device, the losses in the path from the laser output to the input of the device and from the device output to the detector were assumed to be equal, due to the symmetry of the design, giving an estimate of the power seen at the ring which was 0.045 mW and 0.04 mW and a calculated responsivity of approximately 0.09 A/W and 0.11 A/W for design A and design B, respectively. To calculate the dark current of a single device, the neighbouring devices had to be taken into account, since, to save space on the chip, several resonator sensors were designed to be wired together and to share common electrodes. This had the drawback that the dark current from the detectors wired in parallel was summed and the contribution from a single detector could not be directly measured. To estimate the dark current of only the detector under test, the dark current per unit length of each detector was assumed to be equal and the measured current was scaled by the ratio of the length of the detector under test to the combined lengths of all of the detectors that were in parallel. For the devices reported on here, the measured dark current calculated in this manner was 17 nA for the design A device and 30 nA for the design B device. The corresponding Noise Equivalent Power (NEP) for these two designs are 0.19 μ W and 0.27 μ W. Considering that the peak currents for designs A and B devices were 4 μ A and 4.5 μ A, respectively, the highest SNRs that were achieved were 235 and 160, respectively. Table 1 summarizes the characterization results. The relatively low quality factors (Q 's) for these photodetector-sensors, as compared to ring resonators without integrated detectors, is due to the loss introduced by the ion-implantation in the silicon waveguides. As expected, design B has a lower Q than design A due to the loss associated with the highly doped silicon trace crossing the waveguide. Subsequently, for the purposes of this paper, we proceed

with the sensitivity analysis and experiments on the design A device only.

Table 1. Summary of the performances of designs A and B

	Dark Current (nA)	NEP (μW)	SNR	Responsivity (A/W)	Q
Design A	17	0.19	235	0.09	5800
Design B	30	0.27	160	0.11	4500

In order to characterize the sensitivity of the photodetector-sensor, the optical and electrical responses were measured simultaneously while the resonator was exposed to various salt solution concentrations. From the optical responses of the ring resonator in water, the Q for the design A device was measured to be approximately 4400; and the extinction ratio was measured to be approximately 30 dB. The measurements were repeated several times for each concentration. The location of the resonant peak is the important information for sensitivity measurements, which can also be found by curve fitting the response spectra. Figure 6(a) shows how the normalized electrical response shifts as a function of wavelength as we change the concentration of the solution. As we can see, the changing concentration of the reagent results in a resonance wavelength shift of the ring resonator. Figure 6(b) shows the resonance wavelength shift plotted against the salt solution's change in refractive index for various concentrations; the slope of the linear best fit gives a sensitivity (S) of approximately 30 nm/RIU. Based on measured values of Q and S , the intrinsic limit-of-detection (iLoD) for this sensor is 1.1×10^{-2} RIU. Considering that 75% of the ring is exposed to the solution, and that the simulated sensitivity for this waveguide is 40 nm/RIU, this yields a predicted sensitivity of 30 nm/RIU, in agreement with our experiments.

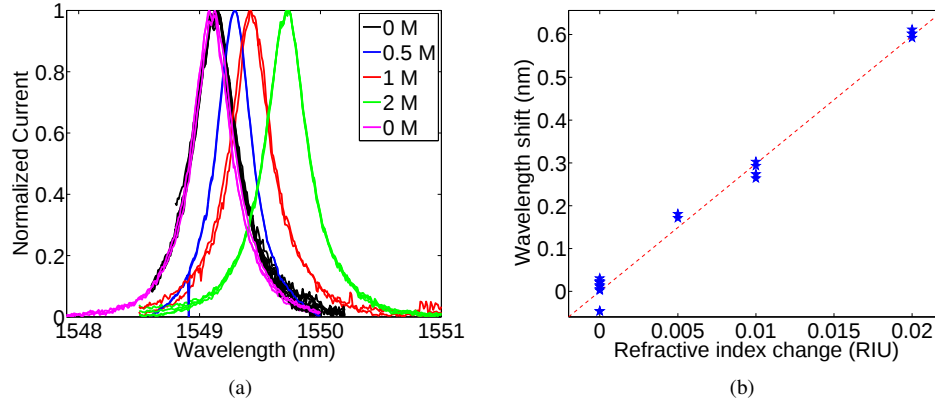


Fig. 6. a) Normalized electrical response of photodetector-sensor design A when exposed to various salt solution concentrations. b) Resonant wavelength shift as a function of the change in refractive index of the solution, for design A. The slope of the best fit line, defined as the sensitivity, is approximately 30 nm/RIU.

6. Analysis and discussion

In this paper, we have investigated and demonstrated the integration of a resonator sensor and detector into one device. If, instead, the resonance-enhanced defect-mediated photodetector

and the resonator sensor were designed as two separated, cascaded devices, then, the resonant wavelengths of the two would need to be matched and therefore, a wavelength locking mechanism would be required. However, having them as a single photodetector-sensor overcomes this wavelength mismatch issue and thus reduces the complexity of making measurements.

6.1. Performance tradeoffs

The various advantages of using this novel approach for easier and cost-effective realization of a system on chip come with the tradeoff that some of the sensing performance is sacrificed. Researchers developing defect-mediated photodetectors have investigated the optimum defect concentration ($N_t = 2 \times 10^{17}$ to $2 \times 10^{18} \text{ cm}^{-3}$), and consequently the optimum ion implantation dosage, for maximum efficiency and detector responsivity in various structures [24,29,30]. Although these defect-mediated detectors have been shown to have performances well comparable with and sometimes better than Ge detectors [32], the integration of the detector into the sensor reduces the Q factor of the resonator (by contributing extra loss on the order of 20 to 30 dB/cm depending on the defect concentration [29,30,32]) which, in turn, degrades the performance of the sensor. We can quantify this reduction in performance by comparing the performance of our integrated photodetector-sensor to a hypothetical device fabricated using the same fabrication process without having an integrated detector. If we were to design this sensor without an integrated detector, we could have designed the resonator to have 100% of its surface exposed to the solution, which would increase the sensor's sensitivity to 40 nm/RIU. In addition, the introduction of defects in the waveguide degraded the Q by about 1000 on average (observed experimentally and confirmed through calculations). Therefore without the defects, the Q would be about about 5400 which would improve the iLoD from 1.1×10^{-2} (for the integrated detector) to 0.7×10^{-2} RIU. In exchange for this slightly lowered performance, the integrated device offers a host of advantages including reduced system footprint and less expensive fabrication process; furthermore, the iLoD of the integrated device could potentially be improved by optimizing the device geometries.

6.2. Discussion and analysis of device performance for sensing applications

Integrated photodetector-sensors, like the one presented here, can be useful for many sensing applications, including the detection of aqueous solutions of changing concentration, gas mixtures, or molecules adsorbed to the surface of the sensor. We have previously studied the SOI-based resonator sensors and have demonstrated both their bulk refractive index sensitivity and surface biosensing capabilities through experiments and simulations [1,2,10,11,33].

For applications which require detecting the concentration of specific molecules in the solution, a change in concentration results in a change in refractive index of the solution. In this application, the sensor's sensitivity to the concentration of specific molecules can be calculated from the sensor's bulk refractive index. For example, the relationship between glucose concentration and RI unit change is described as $\Delta n_{\text{glucose}}(\lambda) = 2.7 \times 10^{-5} \Delta C_{\text{glucose}} [\text{mM}]$ [11,34]. This translates to a sensitivity of approximately 810 pm/M for a device with sensitivity of 30 nm/RIU. Using the same method, the sensitivity for other aqueous solutions and gas mixtures can also be determined.

We have previously demonstrated the capability of our SOI-based resonator sensors in responding to the adsorbed molecular layers using standard sandwich assays [1,10,33]. The experimental results from these previous demonstrations agree with the simulations. In addition, we have confirmed that sensors offering impressive bulk refractive index sensitivities and iLoDs have also performed well in the surface sensing experiments. In the integrated device, we have observed a sensitivity of 30 nm/RIU which is 75% of the simulated value of 40 nm/RIU. Considering that only 75% of the resonator is exposed to the solutions, this shows that our

device's performance agrees with simulations. Based on the agreement between previous experiments and simulations, we expect that we can reliably predict the performance of devices in other applications such as biosensing. For example, if there is a bimolecular add-layer, with refractive index of 1.48 (to mimic proteins), the sensitivity of the non-integrated device (with full surface exposed) to the add-layer thickness is 0.12 nm/nm based on simulations. This sensitivity relation is approximately linear for small add-layer thicknesses of 0 to 30 nm. However, since 75% of the resonator is exposed in the integrated case, we expect a sensitivity of about 90 pm/nm to the thickness of add-layer with refractive index of 1.48.

The integrated device offers many advantages for potential system on chip applications as well as CMOS integration with a small footprint and less expensive mass fabrication process; these advantages might outweigh the sacrificed performance for many applications (particularly low-cost applications or those requiring disposable chips). In addition, the sensing performance of the device could be improved by optimizing the geometries and/or using TM polarization that can potentially provide higher sensitivity [6,35].

7. Summary and conclusion

To avoid using expensive hybrid fabrication of photodetectors for resonator biosensors and system integration on SOI, e.g., using III-V materials or germanium, we proposed the use of a defect-mediated photodetector-sensor. The absorption of the photodetectors were enhanced using resonant structures, giving improved responsivity and SNR, compared to a linear detector. Resonant structures were also shown to be promising biosensors. In this paper, we proposed and demonstrated a novel approach to integrate a sensor with a photodetector by integrating, as a single device, our well-developed resonator-based sensors with defect-mediated photodetectors.

The single-device integration of a photodetector-sensor was investigated by implementing two designs. In each design, the photodetector-sensor could interact with the target analyte through a window in its oxide cladding, where the cladding otherwise provides isolation for the electrical contacts made with the photodetector. In one of these designs, metal routing had to pass over the coupling region, resulting in only a portion of the total length of the photodetector-sensor being exposed to the analyte; whereas in the other design, highly doped silicon was used to route the electrical signal, allowing the full length of photodetector-sensor to be exposed. In the second design, in order to make contact to the inside of the ring, a portion of the highly doped silicon had to pass through the resonator waveguide. This resulted in additional loss in the resonator, and, therefore, a relatively lower Q for the second design. Hence, we performed sensitivity analysis on the design with the higher Q . Our photodetector-sensor had a measured responsivity of 0.09 A/W, sensitivity of 30 nm/RIU, quality factor of 4400 (in water), and extinction ratio of 30 dB.

The integration of a sensor with a photodetector removes the need for an external photodetector to interrogate the sensor and provides a significant step toward the development of CMOS compatible, SOI-based systems on a chip. Implementing a defect-mediated photodetector in a resonator-based sensor (photodetector-sensor) allows for reductions in the device footprint, optical loss, complexity, and cost.

There are two factors to consider when choosing the defect concentration for a detector-sensor: 1) a defect concentration that is too high reduces the Q and iLoD (assuming constant S); therefore degrading the sensor's performance [1,2]. 2) a defect concentration that is too low degrades the detector's performance by reducing the responsivity and SNR of the detector [30], while there is an optimal point for implant concentration and detector's performance [30]. These opposing trends impact the overall system's performance.

Acknowledgments

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