Label-free silicon photonic biosensors for use in clinical diagnostics

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ABSTRACT

Silicon photonics is poised to revolutionize biosensing applications, specifically in medical diagnostics. Optical sensors can be designed to improve clinically-relevant diagnostic assays and be functionalized to capture and detect target biomarkers of interest. There are various approaches to designing these sensors - improving the devices’ performance, increasing the interaction of light with the analyte, and matching the characteristics of the biomolecules by using architectures that complement the biosensing application. Using e-beam lithography and standard foundry processes, we have investigated Transverse Magnetic (TM) and Transverse Electric (TE) disk and ring resonators. TM devices hold the potential for higher sensitivity and large-particle sensing capabilities due to the increased penetration distance of light into the analyte. In addition, devices such as slot waveguide Bragg grating sensors have shown high sensitivities and high quality factors and may present advantages for specific biosensing applications. These devices have been investigated for wavelengths around $\lambda = 1550$ nm (conventional wavelength window in fiber-optic communication) and $\lambda = 1220$ nm, where the water absorption is greatly decreased, offering improved limits of detection. Using reversibly bonded PDMS microfluidic flow cells, the performance and bio-detection capabilities of these devices were characterized. Comparing binding performance across these devices will help validate architectures suitable for biological applications. The most promising sensors for each application will then be identified for further study and development. This paper will discuss the sensors’ comparative advantages for different applications in biosensing and provide an outlook for future work in this field.

1. INTRODUCTION

Just as electronic integration has dramatically changed everyday life (e.g. with tablets, smart phones, laptops and all kinds of electronic devices), photonics integration is quickly evolving and revolutionizing several fields, from optical intra- and inter-chip connections\textsuperscript{1} and high speed telecommunications\textsuperscript{2}, to environmental monitoring\textsuperscript{3} and healthcare.\textsuperscript{4} The fabrication of silicon photonics chips could exploit current CMOS foundries, resulting in a great number of potential advantages, including integration with control electronics, availability of electro-optic devices, and lowered cost facilitated by large scale production and leveraging of existing electronic facilities. Nonetheless, silicon photonics has not yet been chosen as the standard for photonics integration; several platforms have been proposed and developed so far (e.g. III-V semiconductor\textsuperscript{5}, polymers\textsuperscript{6}, etc.), and this technology still needs to address some issues (e.g. the performance of integrated detectors, monolithic light source integration\textsuperscript{7}).

One of the most promising applications of silicon photonics is in the field of biosensing. Sensitive, reliable, and inexpensive silicon phbotonic biosensors integrated with microfluidics and electronics could lead to the development of integrated electro-optic microfluidic chips (Lab on a chip, LOC). These chips will eventually be able to perform completely automated biological analysis for clinical diagnostics and present numerous advantages, such as small sample volume requirement, portability, and ease of use.

The core component of a biosensing LOC is the sensing device, which allows the detection and quantification of target molecules. Several optical sensing mechanisms have been proposed and developed so far\textsuperscript{8,9}; in particular surface plasmon resonance has been exploited in well-developed commercial devices\textsuperscript{10}. Those commercial devices present some drawbacks and only a very small part of the device is actually integrated on a chip. One way to improve current sensing mechanisms is using optical resonators that offer smaller footprints\textsuperscript{11} while keeping
high sensitivity (comparable with commercial devices\textsuperscript{12}). The resonant condition provides enhanced light -
functionalized area interaction because the light travels multiple times and resonates within the sensing volume,
and therefore interacts multiple times with the analyte, causing the sensor response to be amplified.\textsuperscript{13}

Our group is working on characterizing and comparing various types of resonator-based optical sensors. Most
of the silicon photonics chips that are fabricated on a Silicon-on-Insulator (SOI) wafer are waveguide-based
sensors. The high refractive index of silicon confines the light in planar nanoscaled waveguides. Their basic
working mechanism is the detection of a change in the effective refractive index for light propagation in the
structure\textsuperscript{14}, due to a variation of analyte concentration (bulk sensing) or binding of molecules on the waveguide
functionalized surfaces (surface sensing). The addition of a resonant structure allows for sensing this refractive
index change by a shift in the resonance wavelength, which is easier and more robust than intensity change
evaluation.

The aim of this paper is to compare the performance of different optical resonator devices (for the purpose
of optimized sensing devices), i.e. rings, disks and slot-waveguide Bragg gratings, that have been designed and
fabricated, using electron beam (e-beam) lithography rapid prototyping as well as standard CMOS foundry
fabrication processes on SOI chips. Due to the high index of Si ($n_{Si}$ is about 3.46 at 1550 nm, while $n_{SiO_2}$ is
about 1.43), the light is well confined in the silicon waveguide, resulting in low propagation and bending losses for
the TE polarization.\textsuperscript{15} Nevertheless, some modal energy propagates in the surrounding media and interacts with the
analyte. The overlap of electric field and analyte is dictated by the shape of the propagating optical mode and
determines the sensitivity of the device. The geometry (ring, disk, Bragg grating, as well as waveguide type and
size) and polarization of light (TM vs TE) can be optimized to design the sensing element for a given analyte and
application. For example, the TM and TE have different distributions of the field in the waveguide and outside
of it, so they can potentially allow for sensing particles with different sizes or distances from the sensor surface.\textsuperscript{16}
The field distribution for the TM mode has more evanescent field extending into the surrounding media (and
thus in the analyte) so it can potentially provide higher sensitivity and it can allow sensing of particles farther
from the waveguide surface, for example attached to long functionalization chemistries.

To compare the performance of the devices, the set of criteria explained in our previous work\textsuperscript{4} is used. They
are briefly as follows:

1. Sensitivity (S) is the measured wavelength shift as the bulk refractive index unit changes [nm/RIU].

$$ S = \frac{\Delta \lambda}{\Delta n_{fluid}} $$

When comparing different wavelength resonators, normalized sensitivity (S') is used:

$$ S' = S \frac{1}{\lambda} = \frac{S}{\Delta \lambda \Delta n_{fluid}} $$

2. The Quality factor (Q) reflects the number of oscillations until the resonating energy decays to 1/e value.
The higher the number of these oscillations, the more light interacts with analyte. High quality factor
means improved minimum detectable wavelength (leading to improved limit of detection).

$$ Q = \frac{\omega r}{dE/dt} \approx \frac{\lambda_{res}}{\Delta \lambda_{3dB}} $$

3. Intrinsic Limit of Detection (ILOD): minimum refractive index unit change that can be detected by the
resonator, not taking the other system components (eg. laser, readout hardware) into account.

$$ \Delta n_{min} = ILOD = \frac{1}{Q S'} $$
4. Sensitivity as a function of surface changes

\[
\Delta \lambda_B = \frac{\lambda_B}{n_g A \rho} \frac{\partial n_{\text{eff}}}{\partial t} dM
\]  

(5)

where \( t \) is the thickness of the molecular layer, \( A \) is the sensing area, and \( \rho \) is the mass density, \( M = \rho A t \), and \( n_{\text{eff}} \) is the effective refractive index seen by the propagating mode.

Using these metrics, we analyze the performance of ring, disk, and Bragg grating resonators fabricated in SOI, and integrated with microfluidics. We also demonstrate preliminary multiplexing of sensors (multiple resonators interrogated using the same input/output waveguide) and a biosensing experiment.

In addition to investigating these proposed devices at around 1550 nm wavelength, which is commonly used for telecommunication applications, some of these devices are under investigation for wavelengths around 1220 nm. At 1220 nm, light absorption from water is strongly decreased with respect to the 1550 nm range, thus increasing the quality factor and improving the limit of detection.

2. METHODS AND MATERIALS

This section will include details of the design and fabrication of the sensors, the experimental setup, and the reagents used for sensor characterization and bio-experiments.

2.1 Sensor Design and Fabrication

To design the parameters of the various sensor types, simulations were conducted using Lumerical’s MODE and FDTD Solutions, as well as analytical modelling in MATLAB.

Optical devices were fabricated on SOI chips, by e-beam at the University of Washington Microfabrication Facility, as well as by deep UV photolithography through the ePIXfab multi-project wafer service at IMEC. Light is coupled into and out of the chip using grating couplers (GCs). The sensors designed for biological applications were all aligned towards the center of a large chip containing multiple designs, to allow for polydimethylsiloxane (PDMS) bonding on chip and for microfluidic channels to run over them and exposure to various reagents. On the chip, PDMS channels are reversibly bonded, in order to allow the flow of liquids and biological fluids on the top of the devices. Sensors are fabricated with reference sensors, which are not subjected to change in analyte concentration or solution (they can be under PDMS, or in a different channel), thus they can allow correction for temperature drift or other effects.

2.2 Experimental Setup

We have improved upon our previously-described experimental setup to facilitate automatic measurements. Our current setup is capable of automatically measuring hundreds of devices within only a few hours.

2.2.1 Automated Measurement

A tunable laser (Agilent 81682A, Agilent Technologies, Inc., USA) with an output range from 1460 nm to 1580 nm is used as optical source. An array of polarization maintaining optical fibers (PLC connections LLC., USA) is used with fiber grating couplers on chip to inject light into the Si waveguides and capture light coming out of the chip. The output light intensity is measured with an optical power sensor (Agilent 81635A, Agilent Technologies, Inc., USA).

A vacuum is used to immobilize the photonic chip onto the motorized stage (Thorlabs, USA). Light coupling efficiency from the fiber array to the Si waveguides and back out from chip is very sensitive to the alignment (rotational and linear alignment) of the fiber array itself to the fiber grating couplers. By monitoring the output power and moving the stage horizontally the alignment is automatically optimized with a MATLAB script by maximizing collected output power.

A MATLAB script is also used to sweep the laser wavelength and to acquire the transmission spectrum. Furthermore, the script is also used to consecutively measure multiple devices on different locations on the same optical chip, known their coordinates. The stage temperature is controlled with a Peltier element and a feedback controller (LDC501, Stanford Research System, USA) with a typical stability of 1-10 mK.
2.2.2 Microfluidic Integration

Custom design and fabrication of the microfluidic channels in poly dimethylsiloxane (PDMS, Sylgard 184, Dow Corning, USA), using soft-lithography procedures, provided the means to expose specific set of devices on SOI chip to the fluidic reagents. The mold masters were fabricated with standard photolithography in SU-8 2075 (MicroChem, USA). Uncured PDMS was then poured onto the molds to a thickness of about 1 cm. Before curing for 2h at 80°C, the PDMS was degassed in a desiccator for 10min to remove unwanted air bubbles. Inlet and outlet holes are punched into the PDMS layer to access the microchannels and to connect to the syringe pump. The width and height of these microchannels are 100 μm and 90 μm respectively. The PDMS fluidic block was then aligned to the devices under study on the SOI substrate using a stereo microscope. The reversible bond formed between the SOI substrate and the PDMS block is strong enough to form a seal to withstand the pressure used to drive the flow during our experiment; however, to minimize the risk of leakage, the fluids were supplied to the channels under negative pressure (the syringe pump was set to withdraw rather than inject). All reagents were supplied to the channels at a flow rate of 10 μL/min.

2.3 Refractive Index Calibration

To characterize the performance and sensitivity of the devices, a set of aqueous solutions of NaCl were used with various concentrations (7 samples in the range of 0 to 2M). Refractive index of these solutions was measured with a refractometer to ensure the accuracy of the characterization.

2.4 Bio-Assay Experimental Reagents

To test the performance and response of the devices to bio-molecules, we used a modified sandwich assay involving well-characterized molecules with high binding affinities including: anti-Streptavidin (antiSA, Vector Labs; Burlingame, CA), streptavidin (SA, Vector Labs; Burlingame, CA), and biotin-BSA (bBSA), which was conjugated per the manufacturer’s instructions using a commercial biotinylation kit (SoluLink; San Diego, CA). During the experiments, the optical stage was thermally controlled to 30°C to minimize thermal drift. Reagents were introduced to the sensor arrays using a reversibly bonded PDMS flow cell and Chemyx Nexus 3000 Syringe.
Pump (Houston, TX) at 10 µL/min. Selected peaks were tracked every 45 seconds using an Agilent 8164A 1550 nm mainframe with tunable laser (Agilent 81682A) with integrated detector (Agilent 81635A). Each sensor was exposed to phosphate-buffered saline (PBS) for at least 20 minutes prior to other reagents to establish an initial signal baseline.

3. SENSOR DESIGNS AND CALIBRATION RESULTS

When dealing with resonator sensors, the sensitivity is determined by the shift of the resonant peak wavelength as the refractive index of the solution changes. A set of aqueous solutions of NaCl, as explained in section 2.3 is used to characterize the sensor devices under investigation.

Resonator sensors with a high quality factor are desirable. A high quality factor of the sensor means improved accuracy of the detection due to the improved minimum detectable wavelength shift. This will consequently improve the Limit of Detection (LOD) based on equation 4.

3.1 Optical Sensor Designs

Optical resonators have shown promises as biosensors due to their longer interaction with the analyte surrounding the resonators. The following resonators have been investigated in our group: disk resonators; slot waveguide ring resonators; strip waveguide Bragg gratings; and slot waveguide Bragg gratings. Figure 2 provides an SEM image of each of these resonators. Also, since they are integrated in waveguides, they often have compact footprint and thus are easily integrated with microfluidic channels.

![SEM images of the devices fabricated through IMEC. a) disk resonators; b) slot waveguide ring resonator; c) strip waveguide Bragg grating; d) slot waveguide Bragg grating.](image)

Figure 2. SEM images of the devices fabricated through IMEC. a) disk resonators; b) slot waveguide ring resonator; c) strip waveguide Bragg grating; d) slot waveguide Bragg grating.

3.2 Disk Resonators

Disk resonators offer potential advantages of improved limits of detection due to lower scattering losses and thus higher quality factors. The nature of disk resonators means that there is only one sidewall surface from which scattering can occur. These reduced losses increase the resonator quality factor and in doing so have the potential to improve the limit of detection as defined in Equation 4. Additionally, disk resonators offer very small device
footprints; this is advantageous for multiplexing (many disks can fit in a small area, and the disks offer a wide free spectral range, FSR), and the sensing surface area is comparatively small. As described in Equation 5, small sensing areas have the potential to lower the minimum detectable analyte mass, because each molecule sensed will have a higher relative effect on the overall sensing field.

Figure 3 shows the schematics of the 10 µm disk resonators and the simulated mode profiles for the first three TE and TM modes.

![Figure 3](image_url)

Figure 3. Mode profiles (TE and TM) for 10 µm disk resonator. a-c are the first to third TE modes; and d-f are the first to third TM modes.

We have investigated 10 µm disk resonators supporting both TE and TM modes. As figure 3 illustrates, the TM modes in our disk geometries have electric fields that penetrate further into the analyte as well as more field traveling in the analyte itself; this leads to more interaction of light with analyte and therefore higher refractive index sensitivity. The results of the sensitivity analysis of the 3 µm disk are shown in figure 4, demonstrating the peak shift and refractive index sensitivity calibration for the two modes present in the disk.

3.3 Strip Waveguide Bragg Grating

Integrated waveguide Bragg gratings are also promising candidates for biosensors. Compared with other resonant structures (e.g. ring or disk), waveguide Bragg gratings usually operate at only one particular wavelength (Bragg wavelength) and thus are not limited to FSR for the maximum range of the peak wavelength shift.

Figure 5 represents a schematic of a strip Bragg grating sensor with phase shift; with inset being the mode profile of the main propagating TE mode. The Bragg gratings are realized with corrugations on the lateral sidewalls of the strip. As light travels through the waveguide, the optical mode experiences periodic modulation of the effective refractive index, and the Bragg condition depends on the grating period and the effective refractive index of the medium.

The phase shift region in the strip Bragg grating sensors constructs a cavity with two Bragg reflector mirrors and induces a resonant peak in the stop band of the Bragg gratings. The Q factor of the resonance can be very high (100,000 in air).
Figure 4. a) Shows how the spectra shifts as the refractive index of the medium changes (using various concentrations of NaCl). b) Sensitivity for the fundamental (black) second (blue) TE modes in 3 µm radius disk resonator. The error bars indicate a 99% confidence interval (within 3 standard deviation). We observe sensitivities of 26 nm/RIU for the fundamental mode and 29 nm/RIU for the second mode.

Figure 5. Schematic of a strip Bragg waveguide and the propagating TE mode in the waveguide cross-section.

Figure 6 shows the experimental results of the sensitivity analysis of strip waveguide Bragg grating sensor. A sensitivity of 59 nm/RIU is measured, which is close to the simulated value of about 55 nm/RIU. The quality factor (Q) of this device is measured to be 27600, which leads to an intrinsic limit of detection of $9.3 \times 10^{-4}$. 
Figure 6. a) Shows how the spectra shifts as the refractive index of the medium changes (using various concentrations of NaCl). b) Shows the sensitivity of the Strip Bragg Sensors. Error bars show experimental results for peak wavelength shift versus refractive index change of the medium. The error bars indicate a 99% confidence interval (within 3 standard deviation). Red line is a linear fit to these points indicating the sensitivity of about 59 nm/RIU.

3.4 Slot Waveguide Ring Resonator
Slot waveguide ring resonator with 300 nm waveguide width and 130 nm slot were fabricated. Quality factors of about 1450 and sensitivities of 263 nm/RIU for a slot waveguide racetrack resonator, with 30 µm radius, were measured. The low Q is partly due to high bending, mode mismatch, and scattering losses. This limited Q suggests that it is necessary to eliminate the bent regions in the slot device in order to obtain high ILODs.

3.5 Slot Waveguide Bragg Grating
To enhance the light-analyte interaction, we also applied the phase-shifted Bragg grating structure in a slot waveguide\textsuperscript{21}, as shown in Figure 7. The electric field of the slot waveguide is concentrated inside the small low-index slot region. This unique property makes the slot waveguide much more sensitive to the surrounding fluidics than conventional strip waveguides that use only weak evanescent field tails. The Bragg gratings are constructed by corrugating the outer sidewalls of the slot waveguide, where the evanescent field decays.

![Figure 7. Schematic of a slot bragg waveguide and the propagating TE mode in the waveguide cross-section.](image)
Figure 8 shows the experimental results of the sensitivity analysis of a slot waveguide Bragg grating sensor. A sensitivity of 340 nm/RIU is measured, which is in excellent agreement with simulation results.

The quality factor (Q) of this device is measured to be about 15000, which leads to a limit of detection of $3.0 \times 10^{-4}$. Compared to the slot waveguide ring resonators, the slot Bragg grating resonators exhibit significantly enhanced Q factors since they do not suffer from the bending and mode mismatch losses as in ring structures.

4. BIOLOGICAL RESULTS

As a first step towards demonstrating the sensor’s biosensing capabilities, we performed a modified “sandwich” assay using streptavidin (SA) as a model protein and its binding partners, biotin and a monoclonal anti-SA antibody. For the purposes of this study, these molecules were selected based on their availability and ease of use. Additionally, it is worth noting that the specific molecular recognition event between SA and biotin is one of the strongest non-covalent bonds known and anti-SA has been used to demonstrate specific binding to immobilized SA on silicon photonic sensors previously.

Figure 9 shows results from the biosensing experiments for a 3 µm radius TE-mode resonant disk and a single-waveguide Bragg interferometer sensor. Wavelength shifts resulting from the molecular binding events for the disk resonator and Bragg sensor are shown in figure 9 (a and b) respectively. Figure 9 (c) illustrates the idealized sequence of molecular interactions resulting in the sandwich assay. These sequential binding interactions correspond to the wavelength shifts shown in regions i, ii, and iii shown in figure 9 (a, b).

After establishing a signal baseline in PBS buffer, biotinylated-BSA (b-BSA) was adsorbed to the oxide of the sensor surface (shown in region i in figure 9 (a, b)). Next, SA was introduced to the functionalized sensor, binding irreversibly to the immobilized b-BSA, resulting in the wavelength shift observed in region ii (figure 9 (a, b)). Finally, Anti-SA bound the capture SA, serving as a final signal amplification step, resulting in the additional resonant peak shift shown in region iii (figure 9 (a, b, and c)).

These binding results are in good agreement with the formation of a multi addlayer biomolecular system consisting of b-BSA, SA and anti-SA. Ideally, the refractive index change (and subsequent wavelength shift) resulting from each additional layer would correlate precisely to the molecule’s mass of each protein. However, steric hindrance due to dense molecular packing limits 1:1 stoichiometries of binding. This, coupled with the exponential decay of the evanescent sensing field, causes the sensor’s response to each addlayer to deviate from the ideal. With those limitations in mind, the sandwich assay clearly demonstrates expected responses to the biological interactions and demonstrates the platform’s suitability for interrogating molecules in other multi-layer biological assays.
Figure 9. Wavelength shift during biosensing demonstration for the 3 µm radius TE-mode disk sensor (a) and single waveguide Bragg interferometer sensor (b). (c) Illustrates reagent sequencing corresponding to regions [i, ii, and iii] in (a) and (b). Region i = Biotinylated Bovine Serum Albumen (b-BSA) (2 mg/mL), ii = streptavidin (SA) (1.8 µM), iii = anti-streptavidin (anti-SA) (125 µg/mL). A PBS-wash preceded and followed the introduction of each reagent in steps i-iii.

5. DISCUSSION AND ANALYSIS

Table 1 presents a summary of geometrical specifics, resonance wavelengths and light polarizations, as well as sensing performance in terms of Q, S, and ILOD (as defined in paragraph 1) of our fabricated SOI sensors. In addition, Figure 10 presents a comparison of the limits of detection and sensitivities of devices summarized in Table 1 to our previously-presented devices and to limits of detection of various sensor configurations reported in literature. The blue line on Figure 10 represents the theoretical limit to the obtainable sensor performance (in water).

Table 1. Summary of the performance and characteristics of our silicon photonics sensors

<table>
<thead>
<tr>
<th>Sensor</th>
<th>Specifics</th>
<th>λ (nm)</th>
<th>Polarization</th>
<th>Q in water</th>
<th>S (nm/RIU)</th>
<th>ILOD (RIU)</th>
</tr>
</thead>
<tbody>
<tr>
<td>Disk</td>
<td>3 µm radius</td>
<td>1527</td>
<td>TE</td>
<td>32300</td>
<td>26</td>
<td>1.8×10⁻³</td>
</tr>
<tr>
<td>Disk</td>
<td>10 µm radius</td>
<td>1512</td>
<td>TE</td>
<td>131000</td>
<td>21</td>
<td>5.5×10⁻⁴</td>
</tr>
<tr>
<td>Disk</td>
<td>10 µm radius</td>
<td>1543</td>
<td>TM</td>
<td>16000</td>
<td>142</td>
<td>6.8×10⁻⁴</td>
</tr>
<tr>
<td>Slot Ring</td>
<td>30 µm radius</td>
<td>1500</td>
<td>TE</td>
<td>1450</td>
<td>263</td>
<td>1.3×10⁻²</td>
</tr>
<tr>
<td>Strip Bragg</td>
<td>0.5 µm width~112 µm length</td>
<td>1517</td>
<td>TE</td>
<td>27600</td>
<td>59</td>
<td>9.3×10⁻⁴</td>
</tr>
<tr>
<td>Slot Bragg</td>
<td>0.5 µm width~112 µm length</td>
<td>1530</td>
<td>TE</td>
<td>15000</td>
<td>340</td>
<td>3.0×10⁻⁴</td>
</tr>
</tbody>
</table>

It is worth noting that the lowest limits of detection (about one order of magnitude less than other sensor configurations) are achieved with the TE and TM modes in the 10 µm radius disks as well as the slot Bragg.
grating resonator sensors. On the other hand, their refractive index sensitivities vary by an order of magnitude (with TM disk and the slot Bragg with better performance), but their different quality factors equalize the limits of detection to the same order of magnitude. Concerning Bragg sensors, they both have a very low limit of detection and high Q, but the sensitivity is about 6 times improved for the configuration with the slot waveguide. This is likely due to an enhanced interaction between the electromagnetic field and light, causing a wider shift of its resonance wavelength. We can observe in Figure 10 that Q factor of slot Bragg is very close to the theoretical limit, so one can conclude that it is mainly limited by the water absorption.

In addition to ILODs, often resonator characteristics can influence the choice of best resonator type for a given application. Depending on the shape and size of the microchannels, number of multiplexed sensors, as well as shape of target molecule, one can determine the best sensor for the particular application. For example, the 3 µm radius disks have the smallest footprint, while bragg sensors have long but very narrow shape; this perhaps makes 3 µm disks better suited for multiplexing applications. Indeed, the ultimate best choice of sensor type will be dictated by the requirements of the application; whether sharp peaks or large peak shifts are preferable, whether it is advantageous to track multiple peaks at the same time. Obviously, the sizes and expected concentrations of the analyte or of the specific target molecule will all play a role in determining the best sensor type. For example, slot waveguides may be better suited for sensing small molecules with low concentrations, because they offer high sensitivity, but they require that molecules can flow also in the slot (that has a cross section of 150 nm × 220 nm), where much of the field is concentrated.

![Figure 10](image-url)

Figure 10. Experimental results for performance and figure of merit (ILOD) of various sensors. All the red filled points are devices developed by our group at UBC/UW; and black hallowed ones represent devices developed in other groups for comparison purposes. The y-axis is the sensitivity normalized to the peak resonant wavelength (equation 2). The blue line is the theoretical limit for 1550 nm in water.

6. CONCLUSION AND FUTURE WORK

In this paper we have discussed the emerging and promising role of integrated optical biosensors, and we described various individual resonator sensors that have been modeled and fabricated by our group. Those sensors have
been characterized and validated using a standard ‘sandwich’ assay. Having variety of sensors characterized have prepared us for performing more meaningful bio-assay experiments using more custom designed sensors to detect essential aspects under study.

Having sensor with small footprints, it would be very easy to have high-throughput multiplexed and simultaneous analysis, with hundreds of sensors each-one ‘tuned’ for a different specific target. All those sensors can be designed and aligned such that they can be easily integrated with microfluidic channels, as well as they can have built-in normalization with integrated reference sensors. Considering that all those sensors have been fabricated with standard SOI chip processes that are CMOS compatible, facilitating integration with on-chip electronics, they are very promising in terms of whole systems integration. Furthermore, our integration of these devices with microfluidics and bioassays represents a step forward towards realizing Lab on Chip systems.

A final remark on the wavelength we have used, that is 1550 nm, commonly used for telecommunications, thus it has been very well characterized and offers several low-cost components. Since our aim is biological sensing, it is worth exploring different wavelength regions, which may offer some advantages or the possibility of gaining multi-wavelength information. For example we are start to investigate 1220 nm as a working wavelength, since it presents a significantly reduced water absorption, which in turn may improve the figure of merit, ILOD. As seen in the work by Chrostowski et al. (figure 8b), the detection limit is lower at around 1220 nm compared to 1550 nm, making the wavelength window of 1.2 to 1.3 µm a more desirable range for sensing applications in the presence of water.

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