Improving the Performance of Silicon Photonic Optical Resonator-based Sensors for Biomedical Applications

by

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Improving the Performance of Silicon Photonic Optical Resonator-based Sensors for Biomedical Applications

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Abstract

Silicon photonic biosensors show great potential for applications in medical diagnostics and healthcare services. Near-infrared transparency and high refractive index of silicon allow us to build compact and efficient circuits leveraging CMOS foundries, which provide low-cost mass production and enable the integration of the optoelectronic components on the same chip. Although silicon photonic biosensors have proven performances close to today's gold standard diagnostics, many applications still require higher multiplexing, as well as more sensitive, reliable and quantitative measurements. This dissertation is based on theoretical and experimental studies of silicon photonic sensing architectures in terms of sensor performance improvement and unit-cost reduction.

Specifically, two novel sub-wavelength grating-based (SWG) waveguide configurations are presented to improve the sensitivity. Leveraging the advantage of SWG metamaterials, the substrate-overetch (SOE) and multi-box SWG devices present a largely extended modal size and surface contact area, which gives 10time enhanced sensitivity compared to the conventional devices. In addition, by employing the Bragg grating as the sensing architecture, the multi-box SWG-based grating configuration achieves a lower detection limit compared to the microring resonator (MRR) counterpart and demonstrates the capability for monitoring small molecule interactions.

Replacing the laser with a broadband source can provide a lower-cost solution for the optical system. Therefore, two cost-effective broadband light sourcebased sensing implementations are proposed and demonstrated with acceptable sensitivities. The first implementation uses cascaded MRRs for index monitoring, where the analyte variation is converted to the photocurrent change as the readout. The second implementation uses a phase-shifted Bragg grating-based symmetrical Mach-Zehnder interferometer, where the analyte variation maps the intensity change at the resonant wavelength. Furthermore, a system-level integration of active silicon photonic sensors using Fan-Out Wafer-Level-Packaging (FOWLP) is proposed in the dissertation, which can reduce the die size down to 1 mm² while simplifying the microfluidic and optical integration. Leveraging the CMOS foundries and the proposed FOWLP technique, the unit cost of each packaged sensing die can be reduced to several dollars.

Lay Summary

Advanced diagnostic technologies allow people to evaluate processes and events that occurred in vivo. However, in developing countries, medical diagnostics can be expensive and not universal. The key goal of this dissertation is to develop optical sensing architectures to improve diagnostic sensitivity, accuracy, robustness, and economy.

This dissertation has the following contributions: (1) In terms of performance enhancement, two types of sub-wavelength grating-based sensors have been developed, including the substrate-overetch configuration and the multi-box configuration, both of which present an improved sensitivity compared to conventional counterparts. (2) In terms of cost control and reduction, two sensing systems leveraging the low-cost broad-band source have been designed and demonstrated, and a system-level integration technique has been developed for low-cost multiplexed biosensors.

Preface

This dissertation consists of the publications listed in the following paragraph, which came from collaborations with other researchers during my graduate program. Note that only publications contain the work corresponding to the dissertation are shown below. A complete list of publications can be found in Appendix A. Some of the text and figures from publications have been reused in this dissertation with the permission of the publisher.

Journal Publications

 Enxiao Luan, Kashif M. Awan, Karen C. Cheung, and Lukas Chrostowski. "High performance sub-wavelength grating-based resonator sensors with substrate overetch." Optics Letters 44, no. 24 (2019): 5981-5984.

I am the main contributor to this publication. I conceived the idea, designed and simulated the sensing device, conducted optical and biosensing measurements, and wrote the manuscript. Kashif M. Awan provided cleanroom fabrication and optimization ideas. Karen C. Cheung and Lukas Chrostowski supervised the project. Parts of this publication have been used in Chapter 2.

 Enxiao Luan, Han Yun, Minglei Ma, Daniel M. Ratner, Karen C. Cheung, and Lukas Chrostowski. "Label-free biosensing with a multi-box subwavelength phase-shifted Bragg grating waveguide." Biomedical Optics Express 10, no. 9 (2019): 4825-4838.

I am the main contributor to this publication. I conceived the idea, designed the sensing device, conducted optical and biosensing measurements, and wrote the manuscript. Han Yun and Minglei Ma provided methods and technical support for device modelling and simulation. Daniel M. Ratner provided the knowledge for surface chemistry and optimized the biosensing experiments in this project. Karen C. Cheung and Lukas Chrostowski supervised the project. Parts of this publication have been used in Chapter 3.

3) Loic Laplatine*, Enxiao Luan*, Karen C. Cheung, Daniel M. Ratner, Yonathan Dattner, and Lukas Chrostowski. "System-level integration of active silicon photonic biosensors using Fan-Out Wafer-Level-Packaging for low cost and multiplexed point-of-care diagnostic testing." Sensors and Actuators B: Chemical 273 (2018): 1610-1617.

*These authors contributed equally to this work. Loic Laplatine conceived the idea and wrote the manuscript. We built up the sensing setup and optimized the cleanroom fabrication of the proposed device together. I did the cleanroom sensor chip post-fabrication, biosensing performance characterizations, and data analysis. Karen C. Cheung, Daniel M. Ratner, Yonathan Dattner, and Lukas Chrostowski supervised the project. Parts of this publication have been used in Chapter 5.

 Enxiao Luan, Hossam Shoman, Daniel M. Ratner, Karen C. Cheung, and Lukas Chrostowski. "Silicon photonic biosensors using label-free detection." Sensors 18, no. 10 (2018): 3519 (invited).

I am the main contributor to this invited literature review. Hossam Shoman composed part of the manuscript, mainly on the optoelectronic integration. Daniel M. Ratner supervised the manuscript in the aspect of biosensing. Karen C. Cheung and Lukas Chrostowski supervised the project. Parts of this publication have been used in Chapter 1.

5) Enxiao Luan, Han Yun, Loic Laplatine, Yonathan Dattner, Daniel M. Ratner, Karen C. Cheung, and Lukas Chrostowski. "Enhanced sensitivity of subwavelength multibox waveguide microring resonator label-free biosensors." IEEE Journal of Selected Topics in Quantum Electronics 25, no. 3 (2018): 1-11 (invited).

I am the main contributor to this publication. Lukas Chrostowski and I conceived the idea. I conducted the device design and measurement, and wrote the manuscript. Han Yun and Loic Laplatine provided methods and technical support for device modelling and simulation. Yonathan Dattner, Daniel M. Ratner, Karen C. Cheung, and Lukas Chrostowski supervised the project. Parts of this publication have been used in Chapter 3.

Conference Proceedings

 Enxiao Luan, Valentina Donzella, Karen C. Cheung, and Lukas Chrostowski. "Advances in Silicon Photonic Sensors Using Sub-Wavelength Gratings." In 2019 24th OptoElectronics and Communications Conference (OECC) and 2019 International Conference on Photonics in Switching and Computing (PSC), pp. 1-3. IEEE, 2019.

I am the main contributor to this publication. Valentina Donzella provided parts of experiment results. Karen C. Cheung, and Lukas Chrostowski supervised the project.

2) Leanne Dias, Enxiao Luan, Hossam Shoman, Hasitha Jayatilleka, Sudip Shekhar, Lukas Chrostowski, and Nicolas A. F. Jaeger. "Cost-effective, CMOS-compatible, label-free biosensors using doped silicon detectors and a broadband source." In CLEO: Applications and Technology, pp. ATu4K-5. Optical Society of America, 2019.

The idea of the project came from Nicolas A. F. Jaeger, which is an extension work from Hossam Shoman and Hasitha Jayatilleka. Leanne Dias, Hossam Shoman and I composed the manuscript. I contributed to the chip packaging and microfluidics integration for this project. Hossam Shoman provided technical support for the electronic measurements and system control. Leanne Dias and I did the biosensing experimental characterization and data analysis. Sudip Shekhar, Lukas Chrostowski, and Nicolas A. F. Jaeger supervised the project. Parts of this publication have been used in Chapter 4.

 Enxiao Luan, Loic Laplatine, Jonas Flueckiger, Osama Al'Mrayat, Daniel M. Ratner, Karen C. Cheung, and Lukas Chrostowski. "System-Level Integrated Active Silicon Photonic Biosensor for Detecting Small Molecule Interactions." In 2018 IEEE 15th International Conference on Group IV Photonics (GFP), pp. 1-2. IEEE, 2018.

I am the main contributor to this publication. This project is based on the former work from Loic Laplatine and me. Jonas Fluekiger provided the technical support for the sensing setup building. Osama Al'Mrayat provided the simulation and optimization for the sensing architecture. Daniel M. Ratner, Karen C. Cheung, and Lukas Chrostowski supervised the project.

4) Enxiao Luan, Han Yun, Loic Laplatine, Jonas Flueckiger, Yonathan Dattner, Daniel M. Ratner, Karen C. Cheung, and Lukas Chrostowski. "Subwavelength multi-box waveguide-based label-free sensors." In Integrated Optics: Devices, Materials, and Technologies XXII, vol. 10535, p. 105350H. International Society for Optics and Photonics, 2018.

I am the main contributor to this publication. This project is supplementary research for the former work from me. I designed the sensing device, conducted optical and biosensing measurements, and wrote the manuscript. Han Yun and Loic Laplatine provided methods and technical support for device modelling and simulation. Yonathan Dattner, Daniel M. Ratner, Karen C. Cheung, and Lukas Chrostowski supervised the project. Parts of this publication have been used in Chapter 3.

5) Loic Laplatine, Osama Al'Mrayat, Enxiao Luan, Carter Fang, Shayan Rezaiezaden, Daniel M. Ratner, Karen C. Cheung, Yonathan Dattner, and Lukas Chrostowski. "System-level integration of active silicon photonic biosensors." In Microfluidics, BioMEMS, and Medical Microsystems XV, vol. 10061, p. 100610I. International Society for Optics and Photonics, 2017.

The idea of the project came from Loic Laplatine and Lukas Chrostowski. Osama Al'Mrayat provided the simulation and optimization of the sensing architecture. I was involved in the chip design and the cleanroom fabrication of the device. Carter Fang and Shayan Rezaiezadeh built up the sensing setup and composed the control script. Daniel M. Ratner, Karen Cheung, Yonathan Dattner, and Lukas Chrostowski supervised the project. Parts of this publication have been used in Chapter 5.

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Glossary

- BOX Buried oxide
- BSA Bovine Serum Albumin
- **CMOS** Complementary metal-oxide semiconductor
- **DL** Detection limit
- **DUV** Deep ultraviolet
- EBL Electron beam lithography
- ELISA Enzyme-linked immunosorbent assay
- ER Extinction ratio
- FDTD Finite-Difference Time-Domain
- FIB Focused iron beam
- FOWLP Fan-Out Wafer-Level Packaging
- FP Fabry-Perot
- FSR Free spectral range
- FWHM Full width at half maximum
- GC Grating coupler
- IPA Isopropyl alcohol

- I/O Input/Output
- **IVD** In vitro diagnostics
- LOC Lab-on-a-chip
- MPW Multi-project wafer
- MRR Microring resonator
- MW Molecular weight
- MZI Mach-Zehnder interferometer
- PBS Phosphate-buffered saline
- PD Photodetector
- PDMS Polydimethylsiloxane
- PHC Photonic crystal
- PIC Photonic integrated circuit
- PMMA Poly(methyl methacrylate)
- POC Point-of-care
- **PSBG** Phase-shifted Bragg grating
- **RI** Refractive index
- **RIU** Refractive index unit
- SA Streptavidin
- SEM Scanning electron microscopy
- SPR Surface plasmon resonance
- SOA Semiconductor optical amplifiers
- SOE Substrate overetch

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- SOI Silicon-on-insulator
- **SWG** Sub-wavlength grating
- TIA Transimpedance amplifier
- TE Transverse electric
- TM Transverse magnetic
- **TOC** Thermo-optic coefficient
- WG Waveguide
- YI Young interferometer

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Chapter 1

Introduction¹

1.1 Motivation

Medical diagnostics have come to play a critical role in healthcare by providing early detection and diagnosis of disease [1], improving timely and appropriate care [2], protecting the safety of medical products such as blood for transfusion [3], and reducing healthcare costs [4]. Most diagnostic systems have been designed to meet the requirements of well-funded clinical laboratories in highly regulated environments, but do not address the need of the majority of patients and caretakers in the developing world with inadequate healthcare facilities and clinical laboratories [5]. For instance, the enzyme-linked immunosorbent assay (ELISA), which has been the gold-standard method in biomarker detection and validated for more than 40 years, can obtain an ultra-low detection limit (~1 pM) [6]. However, this method is based on a label-based approach that delays results, adds to costs due to specialized reagent requirements, and needs complex micro-evaluations using large, automated analyzers. Therefore, highly sensitive, fast and economical techniques of analysis are desired for both developing and developed countries for point-ofcare (POC) diagnostic applications to improve access to cost-effective healthcare technologies.

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The development of practical biosensors is one of the most promising approaches to satisfy the growing demand for effective medical diagnostic technologies [7]. Since the first oxygen electrode biosensor demonstrated by Clark in 1956 [8], scientists and engineers have made significant progress in the field of biosensing techniques, which has subsequently been adopted into clinical practice. By 2020, the global biosensors market size is anticipated to reach USD 21.17 billion, among which optical biosensors are identified as the most lucrative technology segment. This represents just a fraction of the estimated USD 72 billion worldwide markets for in vitro diagnostics (IVD). There are a variety of techniques that have been successfully employed for optical measurements, such as emission, absorption, fluorescence, refractometry, and polarimetry [9]. Evanescent field detection is the primary detection principle of many optical biosensors [9]. Due to the sensitivity to changes in the local refractive index (RI) within the evanescent field surrounding the device, evanescent field biosensors such as Surface Plasmon Resonance (SPR) or planar waveguide-based sensors have attracted growing interest for sensitive, real-time, and label-free biomolecular detection [10].

Surface plasmon resonance (SPR) is a physical phenomenon that describes the photons energy absorption at a glass/metal interface when incident light enters the interface of two media with different refractive indices at a critical angle. The incident photon energy is absorbed when the photon frequency matches the natural oscillation of surface electrons and causes resonance, thus making the reflected light intensity attenuation within a certain angle. Any molecules adsorbing at the metal surface changes resonant conditions and alters the resonance angle, which can be applied for analytes detection and quantification. Although SPR has been successfully applied to solve various analytical tasks for more than 30 years and exploited in well-developed commercial devices with impressing sensor performance down to 10^{-7} RIU [11], several drawbacks still exist that limit its potential beyond research settings. Due to the complexity of the plasmonic systems integration, SPR has not been realized as a highly-parallelized, portable, low-cost clinical assay [12]. Because of the exponential decay of the surface plasmon from the gold surface, the sensing range of SPR is limited, which is poorly suited for large-size targets, such as bacteria and cells. These targets usually place the majority of the index change outside of the range of the evanescent field [12]. In addition, the relevant path lengths that light can interact with the analyte in the SPR is only twice of the thin metal layer (~ 10 nm).

Silicon photonic integrated circuits (PICs) technology is one of the most promising solutions for the next-generation evanescent field sensors. Due to the compatibility with complementary metal-oxide semiconductor (CMOS) foundry processes, silicon PICs can be manufactured with great efficiency at high volume. Moreover, the high refractive index contrast between silicon and silicon dioxide, or other surrounding media, enables the development of miniaturized compact sensing devices, with the additional possibility of fabricating multiple sensors on one single chip [9]. Meanwhile, silicon photonics are excellent transducers for continuous and quantitative label-free biosensing [13, 14], which can directly respond to affinity interactions between analyte and receptor molecules in real-time. Hence, numerous silicon photonic sensing devices, such as Mach-Zehnder interferometers (MZIs) [15, 16], microring resonators (MRRs) [17, 18], microdisk resonators [19, 20], Bragg grating resonators [21, 22], and one-dimensional (1D) or two-dimensional (2D) photonic crystals (PHCs) [23, 24] have been developed over the past decades for biosensing diagnostic applications. Compared to SPR, light can travel in planar resonators with thousand times of roundtrips, which provides a radically increased interaction length on the order of meters, even the ring is only 20 µm in radius.

It has been reported that silicon photonic sensors had achieved sensitivities close to the clinical relevance sensing in complex media [25], and plenty of siliconbased architectures had been successfully applied for the detection of cell secretions [26], virus [27], protein biomarkers [10], and nucleic acids successfully [28, 29]. Table 1.1 shows the comparison of the analysis time and the detection limit for different biosensors. Among them, MRR-based silicon photonic sensors show competitive detection limits (in picomoles) and analysis time. However, many clinical diagnostic tests still require a cost-effective sensing system with a lower detection limit and a robust and high sensitivity. The motivation of my research is to develop the laboratory-based sensing system for the use in clinicallyrelevant applications through enhancing the sensor performance and reducing the cost of silicon photonic sensors.

Table 1.1: Fluidic detection limits by different types of biosensors (LFA =lateral flow assay, PSA = prostate specific antigen, IFA = immunofluores-cent assay, IgG = Immunoglobulin G, IL-2 = human cytokine interleukin-2).

Catagory	Description	Detection condition	Analysis time	Detection limit
LFA	PSA test	Clinical serum samples	30 min	15.8 pM [30]
IFA	ELISA	Serum samples	60 min	0.1 pM [6]
SPR	Labelled detection	p53 cDNA samples	120 min	1.4 fM [31]
	Label-free detection	IgG solutions	20 min	0.2 μM [32]
MRR	Labelled detection	IL-2 solutions	45 min	6.5 pM [26]
	Label-free detection	Lectins from Aurelia	20 min	10 pM [33]

1.2 Theory and structures

1.2.1 Evanescent field sensing principle

Leveraging the silicon-on-insulator (SOI) platform, silicon photonic biosensors rely on near-infrared light confined in nanometer-scale silicon wires (known as waveguides) to sense molecular interaction events. The portion of the electrical field of light travelling outside of the waveguide is referred to as the evanescent field, which can interact with the surrounding volume to create an external RI sensitive region (Figure 1.1(a)). When target molecules bind to receptors at the waveguide surface, the accumulation of molecules with a different refractive index changes the external RI and perturbs the evanescent field, which then further influences the behavior of the guided light in the waveguide [25]. By monitoring the coupling and/or propagation properties of the output light, analytes of interest can be detected in real-time (Figure 1.1(b)) [34]. Since the evanescent field decays exponentially with a decay length ranging from a few tens to a few hundreds of nanometers into the bulk medium, the sensing signal of an analyte captured within the decay length shows a significant difference compared to the signal of an analyte floating far away from the surface [14]. Thus, based on the response of the evanescent field sensor, we can distinguish the target molecules immobilized on the surface (surface sensing) from those remaining in bulk solution (bulk sensing), as presented in Figure 1.1(c).



Figure 1.1: Principle of the evanescent field detection for a silicon photonic biosensor. (a) The evanescent field (dashed lines) around the waveguide is sensitive to the RI change caused by biological binding events at the waveguide surface. (b) Optical transmission spectra of the sensor before (blue curve) and after (red curve) the analyte interaction, resulting in a wavelength shift ($\Delta\lambda$). (c) Sensorgrams of the sensor in bulk (blue curve) and surface (red curve), where the signals are recorded as a function of time.

Several figures of merit are widely used for the evaluation of sensor performance, such as selectivity, reproducibility, stability, sensitivity, and resolution (detection limit). Selectivity describes the ability of a sensor to detect a target analyte in a sample containing other admixtures, which is the main consideration for the bioreceptor selection; reproducibility is the ability to generate identical responses for repetitive experimental setups, which provides high reliability and robustness for the signal; stability refers to the degree of susceptibility to ambient disturbances around the sensing system, which can affect the precision and accuracy of the sensor [35]. Sensitivity (*S*) and detection limit (DL) are two performance criteria we would like to focus on in this dissertation since they have a stronger correlation with their sensor geometries. In evanescent field-based sensors, sensitivity is determined by the strength of interactions between matter and the fraction of light in solution or at the surface [14]. According to the status of target molecules, two specific types of sensitivities are defined in biosensing applications: (1) bulk sensitivity (S_{bulk}), which takes into account RI changes of the waveguide entire cladding, and (2) surface sensitivity (S_{surf}), which assesses RI changes within the first few tens to hundreds of nanometers above the surface [25]. For the bulk sensitivity, it is defined as the slope of wavelength (or phase) shift versus the change of refractive index unit (RIU), and the shift is described by [36]:

$$\frac{\Delta\lambda}{\lambda} \text{ (or) } \frac{\Delta\phi}{\phi} = K \times \frac{\Delta n_{\text{fluid}}}{n_{\text{g}}} \times \frac{\partial n_{\text{eff}}}{\partial n_{\text{fluid}}}, \tag{1.1}$$

where λ is the wavelength and ϕ is the phase of the input light, *K* is the sensor structural constant (varies depending on the configuration of the sensor), n_{fluid} is the RI of the analyte, and n_{eff} and n_{g} are the mode's effective and group indices. From Equation 1.1, the wavelength (or phase) shift is mainly contributed by the shift in the solution's RI (Δn_{fluid}), the dispersion (n_{g}) of the material and waveguide, and the mode's effective index change ($\partial n_{\text{eff}}/\partial n_{\text{fluid}}$) caused by the slight change of the mode profile [36]. The bulk sensitivity is defined as:

$$S_{\text{bulk}} = \frac{\Delta \lambda \text{ (or) } \Delta \phi}{\Delta n_{\text{fluid}}}.$$
(1.2)

As for the surface sensitivity, the definition is slightly different from the bulk one by replacing the solution's RI (n_{fluid}) with the thickness of a homogeneous adlayer on the surface (t_{adlayer}). Therefore, the expressions for the wavelength (or phase) shift and surface sensitivity are [37]:

$$\frac{\Delta\lambda}{\lambda} \text{ (or) } \frac{\Delta\phi}{\phi} = K \times \frac{\Delta t_{\text{adlayer}}}{n_{\text{g}}} \times \frac{\partial n_{\text{eff}}}{\partial t_{\text{adlayer}}}, \tag{1.3}$$

$$S_{\rm surf} = \frac{\Delta\lambda \ (\rm or) \ \Delta\phi}{\Delta t_{\rm adlayer}},\tag{1.4}$$

respectively. From Equation 1.3 and 1.4, $\partial n_{\text{eff}}/\partial t_{\text{adlayer}}$ is highly dependent on the refractive index of the adlayer material: a high RI analyte can lead to a significant effective index variation and wavelength shift even with a thin adlayer at the sur-

face. Thus, surface sensitivity is usually defined for a specific molecule of interest and is not suitable for a general comparison among sensors operated with different biosensing assays.

The detection limit (DL) is typically specified as the minimum RI (or smallest mass) change necessary to cause a detectable change in the output signal, and defined as follows:

$$DL = \frac{3\sigma}{S} \tag{1.5}$$

where σ is the system noise floor, and *S* is the bulk or surface sensitivity. Since σ depends on the experimental setup and readout instrumentation, this DL is also regarded as the system detection limit (sDL). For an evanescent field label-free biosensor, DL can be specified in three units: (1) DL in units of refractive index units (RIU) aims to characterize the sensing capability in bulk solution, which offers a rough comparison among different sensors, (2) DL in units of pg/mm² and (3) in units of ng/mL aim to characterize the sensing capability at sensor surface by using surface mass density and sample concentration, respectively [14]. Due to the correlation among these DLs, the sensing capability of optical biosensors based on different bioassays can be investigated and compared.

1.2.2 Interferometer based biosensors

Interferometer-based biosensors constitute one of the most sensitive integratedoptic approaches by combining two very sensitive methods: waveguiding and interferometry techniques [38]. In a conventional interferometric biosensor, the guided light is split by a Y-junction into two single-mode waveguide paths, one of which containing the sample is regarded as a sensing arm, and the other one is used as a reference arm. The evanescent field of the sensing arm interacts with the sample and senses the RI change at the surface, resulting in an optical phase shift. After a certain distance, the beams recombine again and cause a constructive or destructive interference at the output (as shown in Figure 1.2(c)), where the intensity modulation corresponds to the RI difference between sample and reference arms.

Young and Mach-Zehnder interferometers are the most common formats for interferometric sensing techniques [34, 38, 39]. Although both of these interferometers utilize Y-junctions to split the coherent, single-mode and polarized light at



Figure 1.2: Interferometric biosensors. (a) Illustration of a typical Mach-Zehnder interferometer. The light is split into two arms (sensing and reference) and recombined at the output by on-chip Y-junctions. The degree of interference is proportional to the RI variation taking place on the sensing arm. (b) Illustration of a classic Young interferometer. Rather than using Y-junctions to rejoin the split beams, the light is projected from two closely spaced secondary sources onto a CCD camera, resulting in an interference pattern. (c) Measured interferogram of a typical MZI device after normalization by eliminating the insertion loss.

the input, the output recombination of Young interferometers (YIs) is not realized like MZIs (Figure 1.2(a)) by another on-chip Y-junction. Instead, the interference light in YIs is projected on a screen or CCD camera in an off-chip way, as shown in Figure 1.2(b).

In case of an MZI sensor, the output intensity (I_{out}) is a periodically oscillating function of the phase change difference $(\Delta \phi)$ of the beams from two arms with the following expression [40]:

$$I_{\text{out}} = I_{\text{sen}} + I_{\text{ref}} + 2\sqrt{I_{\text{sen}}I_{\text{ref}}}\cos\left(\Delta\phi + \Delta\phi_0\right)$$
(1.6)

where I_{sen} and I_{ref} are the intensity of the light passing through the sensing and

reference arms of the MZI, respectively, and $\Delta \phi_0$ is the initial phase difference due to the unbalance of the two arms. The phase change difference caused by the variation of the effective index (Δn_{eff}) at the wavelength λ is calculated as:

$$\Delta \phi = \frac{2\pi}{\lambda} \Delta n_{\rm eff} L \tag{1.7}$$

where L is the effective detection length of the sensing arm. The sensitivity of interferometric sensors is defined as the change in phase caused by the change in the RIU of the cladding above the sensing arm. According to Equation 1.7, a longer interaction length (L) in the sensing arm can increase the sensitivity [41]. However, due to the cosine-dependent intensity function of the interferometric curve, the intensity response is non-linear: a higher signal change at the quadrature point is observed than the one near the curve extreme of the cosine function. Moreover, false-positive signals occur when input source fluctuations or temperature variations happen, which strongly influence the reliability of the interferometric sensor, especially with long sensing arms [42]. Thus, additional modulation approaches are usually needed to tune the phase difference between the arms for interferometer sensors.

1.2.3 Resonant microcavity based biosensors

Optical microcavity resonators have been investigated as an emerging sensing technology due to their potential for highly-compact sensing arrays. In a microcavity resonator structure, incident light propagating in an input waveguide or tapered fiber is coupled into the microcavity via the evanescent field. Then, coupled light passes through the cavity in the form of whispering gallery modes (WGMs) or circulating waveguide modes with multiple round-trips, resulting in optical interference at specific wavelengths of light, as shown in Figure 1.3(d) by the resonant condition:

$$\lambda = \frac{2\pi r \times n_{\rm eff}}{m} \tag{1.8}$$

where λ is the resonant wavelength, *r* is the radius of the resonator, n_{eff} is the resonator effective refractive index, and *m* is an integer. The positions of resonant peaks are related to the RI near the resonator surface and shift due to the change

of n_{eff} , which can be monitored by scanning the wavelength or by measuring the intensity at a single wavelength.

Unlike interferometric biosensors, the interaction of light and analyte is no longer determined by the length of the sensing waveguide, but rather by the characteristic time of the energy stored inside the resonator, which is evaluated by the quality factor (Q) [14]. Q describes the photon lifetime in the resonator and represents the number of oscillations before the energy has decayed to 37% (1/e). Therefore, Q incorporates the distributed loss of a resonator and is approximated by dividing the resonant wavelength by its full width at half maximum (FWHM) [36]:

$$Q = \omega \frac{\varepsilon}{\partial \varepsilon / \partial t} = \frac{2\pi n_{\rm g} \times 4.34}{\lambda \times \alpha_{\rm (dB/m)}} \approx \frac{\lambda}{\Delta \lambda_{\rm FWHM}}$$
(1.9)

where ω is the resonant frequency, ε is the energy of the resonant mode, n_g is the group index, α is the total distributed loss in the resonator, and $\Delta\lambda_{FWHM}$ is the FWHM bandwidth of the resonance peak. A higher Q indicates that light stays in the resonator longer and interacts more with the analyte. Moreover, White et al. have proved that having a high Q is advantageous in reducing the noise of the sensor (3σ), which further improves the DL [43]. As mentioned before, the DL (or sDL) relies much on the measurement system, including curve fitting methods and limitations from light sources or detectors, which makes it difficult to have an objective comparison between sensors with different assays and experimental systems [44]. As a consequence, intrinsic detection limit (iDL) was introduced as a substitute for resonant sensors, which is only dependent on intrinsic characteristics, i.e., the resonance linewidth, and defined by [45]:

$$iDL = \frac{\lambda}{Q \times S} \tag{1.10}$$

where λ , Q, and S are the sensor resonant wavelength, quality factor, and sensitivity, respectively. By replacing S with S_{bulk} or S_{surf} , the bulk or surface iDL can be represented.



Figure 1.3: Planar resonant microcavity biosensors. (a) Illustration of a conventional MRR sensor. By using a bus waveguide, guided light is coupled into the resonator at a frequency corresponding to the resonant condition. (b) Illustration of a microdisk resonator sensor. (c) Illustration of a microtoroid resonator sensor. (d) Measured transmission spectrum of a conventional MRR device after normalization.

1.2.4 Photonic crystal based biosensors

A photonic crystal (PHC) waveguide consists of periodically repeating arrays of dielectric structures, forming periodic variations in the refractive index. The periodicity is on the order of the optical wavelength and stops a range of wavelengths propagating through the PhC, resulting in a photonic bandgap on the transmission (or reflection) spectrum presented in Figure 1.4(d). By introducing a defect into the PHC structure, a defect mode at a particular wavelength is formed and resonantly confined in the defect region, which leads to a sharp peak within the bandgap. Due to the strong optical confinement, light is concentrated in a minimal volume near the defect, enabling an intense light-matter interaction area. A tiny volume of analytes immobilized surrounding the defect can induce a noticeable shift of the resonance wavelength and provide a measurable response. Hence, in the past ten years, PHC based biosensors are regarded as a promising and novel technology

that has gained much attention [46-48].

The periodicity of a PHC structure can vary from one-dimensional (1D), twodimensional (2D) to three-dimensional (3D). One-dimensional PHCs are the most straightforward architecture analyzed by Lord Rayleigh as early as 1887. These structures consist of different material layers with high and low refractive indices alternatively (Figure 1.4(a)) and are usually fabricated by layer-by-layer deposition, spin coating, or photolithography methods [49]. In 1987, Yablonovitch [50] and John [51] reported the detailed research on PHCs separately, proposing the concept of photonic bandgaps in 2D and 3D structures. 2D and 3D PHCs exhibit their periodicity in two and three spatial directions, as shown in Figure 1.4(b) and 1.4(c), which need complex manufacturing techniques like photolithography, etching, and particle self-assembly, etc [49]. Although the complexity of the manufacturing process of 1D PHC devices is low, a well-collimated beam is usually required for sensing approaches, especially for high Q devices, which needs the sensing area to be relatively large, compared to 2D or 3D ones [52].

1.2.5 Bragg grating based biosensors

The Bragg grating, a fundamental component for the purpose of wavelength selection, has been investigated for use in optical communications, such as filters, semiconductor lasers and fibers for a long time [53], and recently into biosensing applications [21, 54]. Similar to 1D photonic crystals, a Bragg grating is a structure with a periodic modulation of the effective RI in the propagation direction of the optical mode, as shown in Figure 1.5. By alternating the material with different indices or physical dimensions (known as the corrugation) of the waveguide, the desired index modulation is achieved. A reflection of the guided light occurs at each index-changed boundary, as presented in Figure 1.5(a), and the repeated modulations of the effective index multiply the distributed reflection, resulting in a stop band at one specific wavelength in the transmission spectrum, where light is strongly reflected. The center wavelength of the stop band, namely the Bragg wavelength, is given as:

$$\lambda = 2\Lambda n_{\rm eff} \tag{1.11}$$



Figure 1.4: Illustration of photonic crystals in (a) 1D, (b) 2D, (c) 3D conformations. Insert: Schematic representation of each format showing the periodic arrangements, different colors represent materials with different indices. (d) Measured transmission spectrum of a uniform PHC device after normalization.

where Λ is the period, and n_{eff} is the average effective index of Bragg gratings. If a phase-shifted cavity is introduced in the middle of the gratings, as illustrated in Figure 1.5(b), a narrow resonant transmission peak will appear within the stop band [55], which can be utilized for RI change monitoring.

1.2.6 Performance comparison

Figure 1.6 summarizes the simulated transmission spectra of previously described optical configurations in the field of silicon photonic biosensors. As a concept illustration, we only consider the intrinsic losses in each device. As shown in Figure 1.6, MZI (blue curve) and MRR (red curve) sensors present periodic spectra. The spacing between optical wavelengths of two consecutive transmitted optical



Figure 1.5: Bragg grating biosensors. (a) Illustration of two types of Bragg grating devices with side-wall or top gratings. R and T are the grating's reflection and transmission. The 180° arrows represent the numerous reflections throughout the grating. (b) Schematic of a phase-shifted Bragg grating device. A is the period, ΔW is the width of the corrugation, a or b and n_{eff1} or n_{eff2} are the length and the effective index of the high or low index section. (c) Measured transmission spectrum of a phase-shifted Bragg grating device after normalization.

intensity minima is defined as the free spectral range (FSR) and given by:

$$\Delta \lambda_{\rm FSR} = \frac{\lambda^2}{n_{\rm g} \times \Delta L} \tag{1.12}$$

where λ is the wavelength of the light source, n_g is the waveguide group index, and ΔL is the length difference of two arms in the MZI or the perimeter of the MRR. As for the transmission spectrum of the PHC or Bragg grating (yellow curve), due to the existence of the defect or phase-shifted cavity, a sharp FSR-free resonant peak appears in the middle of the stop band with a narrow FWHM corresponding to the high Q. By interrogating the wavelength (phase) shift or intensity change of these peaks in the transmission plots, the RI change caused by the analyte within the evanescent field can be monitored in real-time.



Figure 1.6: Simulated transmission spectra of different optical configurations, including MZI (blue curve), MRR (red curve), defected PHC or phase-shifted Bragg grating (yellow curve) sensors. The optical insertion loss caused by input and output coupling devices has been eliminated. The full width at half maximum (FWHM) indicates the optical wavelength width of the resonant peak at which the transmitted intensity is equal to half (-3 dB) of its maximum value.

As described in Equation 1.1 and 1.3, the sensor structure constant K in a feedback-based (such as MRRs) sensor is 1; whereas in an feedforward-based (such as MZIs) sensor, K equals to $L_1/(L_1 - L_2)$ where L_1 and L_2 are waveguide lengths of sensing and reference arms, respectively. That can be derived in a physical way by introducing a perturbation parameter q into the sensing system [56]. Sensitivity details can be found in Appendix B. The sensitivity is independent of the physical size in an MRR-based sensor, but scales with the length ratio between the sensing arm and arm difference in an MZI-based counterpart, as presented below:

$$S_{\rm MRR} = \frac{\Delta \lambda_{\rm MRR}}{\Delta n_{\rm add}} = \frac{\lambda}{n_{\rm g}} \left(\frac{\partial n_{\rm eff}}{\partial n_{\rm add}} \right), \tag{1.13}$$

and

$$S_{\text{MZI}} = \frac{\Delta \lambda_{\text{MZI}}}{\Delta n_{\text{add}}} = \left(\frac{L_1}{L_1 - L_2}\right) \frac{\lambda}{n_{\text{g}}} \left(\frac{\partial n_{\text{eff}}}{\partial n_{\text{add}}}\right) = \frac{2\pi L_1}{\lambda} \left(\frac{\partial n_{\text{eff}}}{\partial n_{\text{add}}}\right) = \frac{\Delta \phi_{\text{MZI}}}{\Delta n_{\text{add}}}.$$
 (1.14)

From Equation 1.13 and 1.14 above, a common factor $\partial n_{\text{eff}}/\partial n_{\text{add}}$ is defined as the waveguide mode sensitivity (S_{wg}), which is determined by the waveguide mode

distribution in cross-section. However, waveguide mode sensitivity only maps the physical changes around the waveguide into effective index variations, which is not a directly measurable quantity [57]. Therefore, the sensing architecture (resonators or interferometers) is used to convert the effective index change to detectable signals (wavelength shift or phase change), by the architecture sensitivity factor, i.e., λ/n_g and $2\pi L_1/\lambda$, respectively.

Generally, compared to other geometries, the MZI-based optical sensor is one of the simplest configurable devices with good sensitivities that scale with the length of the sensing arm. Disadvantages such as large footprint, high-temperature sensitivity, and the need for additional modulation methods hinder the development of on-chip interferometric sensing arrays. Resonator-based sensors, like MRRs, microdisks, PHCs and Bragg gratings, are more suitable for the integrated sensing platform with a high density due to their small sizes. Different from MRRs, PHCs and Bragg gratings have a high Q due to the elimination of bending (mode and radiation) losses, thus an improved iDL, even though their sensitivities are comparable.

1.3 Performance-improving strategies

We outline early and emerging strategies in the development of SOI-based biosensor performance, including the use of new geometries of optical waveguides, and different polarizations or wavelengths of light sources. Furthermore, an overall performance metrics comparison is presented at the end, which includes proposed sensing architectures with or without their performance improved strategies.

1.3.1 Fundamental approaches

Transverse magnetic mode

Due to the large evanescent field component travelling above the waveguide, optical sensors in the quasi-transverse magnetic (TM) mode present an improved sensitivity to that of the quasi-transverse electric (TE) mode at 1.55 μ m in conventional 220 nm-thick SOI waveguides [58, 59]. Figure 1.7 shows the electric field intensity distributions of the TE and TM modes propagating in a 220 \times 500 nm waveguide.



Figure 1.7: Illustration of electric field intensity distributions of the (**a**) TE and (**b**) TM modes in a 200 × 500 nm silicon waveguide at 1550 nm wavelengths. The Si waveguide core ($n_{\text{eff}} = 3.47$) is exposed to the surrounding medium with a refractive index of 1.33 above a 2 µm thick buried oxide layer (BOX) with a refractive index of 1.44.

Most of the field intensity is above and beneath the waveguide core (in the cladding and substrate) in the TM mode, offering a higher light-matter interaction strength. Moreover, the TM mode also experiences less scattering loss, which is usually caused by sidewall roughness, compared to the TE mode [37]. Because of these unique properties of TM mode based waveguides, a large number of evanescent field biosensors have been attempted in the TM mode for higher susceptibility to RI changes.

Slot Waveguides

A slot-waveguide device consists of two high index rails separated by a low index slot [60]. Because of the high concentration of the electric field intensity within the slot, slot-waveguide based structures stand out for the potential to enhance sensitivity for optical biosensors. As presented in Figure 1.8(a), light is strongly confined in the slot region. Thus, compared to conventional waveguides, a stronger lightmatter interaction can be obtained in this region, leading to improved sensitivity. Also, slot-waveguide based structures are also CMOS compatible which enables miniaturization and integration for a lab-on-a-chip platform with low cost [41, 61].

Thinner Waveguides

Using thinner waveguides can lead to lower optical confinement of the guided mode, resulting in deeper penetration of the evanescent field into the surrounding medium, as seen in Figure 1.8(b). Thus, more field overlap with biomolecules at the waveguide surface is achieved. Moreover, due to the index of the water cladding decreasing with rising temperature, which is opposite to the Si core and SiO₂ substrate materials, ultra-thin TE MRR sensors show increased stability in the presence of temperature variations as compared to the traditional 220 nm thick sensors [62].

Suspended Waveguides

Another method to enhance the overlap between the evanescent field and analyte is introducing suspended waveguides, by replacing the BOX substrate with lowerindex materials (e.g., air and water). Due to the suspended configuration, sensing architectures expose more light into the surrounding media and improve the sensitivity. Furthermore, combined with the unique distribution advantages of the TM mode, sensor performance can be further optimized.

1310 nm Light Sources

For label-free biosensing, one way to improve the limits of detection of silicon photonic sensors for medical diagnostic applications is enhancing the intrinsic sensor performance [37]. According to Equation 1.10, iDL shows a reciprocal relation to its Q and S. Thus, having a large Q or sensitivity value can effectively improve the iDL. The Q can be interpreted as the total distributed loss of the device based on Equation 1.9, and the loss originates from waveguide scattering, material absorption (waveguide and analyte), waveguide radiation, mode mismatch, etc [36]. Among them, water absorption is the predominant loss for silicon photonic biosensors at 1550 nm wavelengths since many analytes of interest are found in aqueous solutions. Kou et al. observed that water absorption is approximately 10 times lower around 1310 nm wavelengths compared to 1550 nm ones [63]. By assuming an ideal Fabry-Perot cavity with the light travelling entirely in the water, where no other loss mechanism exists, a fundamental limit for water-based sensors was



Figure 1.8: (a) Cross-section of the electric field intensity distribution of a slot-waveguide immersed in water. (b) Electric field intensity distributions of a TE mode for 90, 150 and 220 nm thick silicon cores. (c) Fundamental DL plots for water-based sensors at 1310 and 1550 nm wavelengths. Highest predicted DL for water absorption limited sensing is presented (blue line). Waveguide scattering is added and assumed to contribute 5 dB/cm loss at 1550 nm, and scale as $1/\lambda^4$ at other wavelengths (green line). Finally, the sDL is shown (red line) with a wavelength readout precision 100-fold better than the resonator linewidth.

calculated, showing a intrinsic limit of detection of 2.4×10^{-4} RIU at 1550 nm and 2.4×10^{-5} RIU at 1310 nm, respectively in Figure 1.8(c) [36].

1.3.2 Advanced approaches

Sub-wavelength grating waveguides

A novel and appealing strategy, which allows customizing optical properties by varying the waveguide geometry, is using sub-wavelength gratings (SWG) [64]. Since the first demonstrations of an optical waveguide with an SWG metamaterial core by the National Research Council of Canada (NRC) in 2006 [65–67], SWG waveguides have attracted intense research interest due to their unique potentials to control light propagation in planar waveguides, and been considered to be critical components for developing the next generation of optical communica-



Figure 1.9: SWG waveguide geometry and simulation results. (a) Schematic of an SWG waveguide. *W* is the waveguide width, *t* is the thickness, Λ is the SWG period, and η is the duty cycle which determines the length of Si blocks. n_1 and n_2 represent high and low refractive indices. (b) The top and cross-sectional views of the electric field intensity distribution of an SWG waveguide. The cross-sections are in the middle of the Si block and gap, respectively.

tion, biomedical, quantum and sensing technologies [68, 69]. Although similar to Bragg gratings, SWG waveguides also consist of the periodic structure of their core, the period (Λ) is much smaller than the Bragg condition, i.e., $\Lambda \ll \lambda/(2n_{\text{eff}})$. Thus, a true lossless mode is supported in SWG waveguides because the reflection and diffraction effects are suppressed [70]. The SWG waveguide core is commonly fabricated by interleaving the high index block (n_1) with low index materials (n_2), such as SiO₂, SU-8, air or water, as one period (a few hundred nanometers in length), as shown in Figure 1.9(a). By having a reduced mode effective index step, the guided light propagates in SWG waveguides similar to the one in conventional waveguides but with a large extended modal area, which releases more optical mode into the evanescent field. Moreover, as shown in Figure 1.9(b), most of the light is concentrated in the low-index region, which offers direct light-matter contact. Thus, compared to the conventional waveguide, the sensing performance of an SWG waveguide-based biosensor is highly enhanced.

Vernier Effect Based Systems

The Vernier effect is a method commonly used in calipers and barometers to enhance the accuracy of instrument measurements by overlapping two scales with different periods, of which one slides along the other one. The overlap between lines of the two scales is used to perform the measurement. In the past few years, Vernier-principle based sensors have been investigated in the SOI platform by cascading two or more optical devices with different FSR values, where one has the upper cladding removed and represents the RI sensor (as seen in Figure 1.10(a)). Due to the different FSRs between the sensing and reference (filter) devices, a spectral response with a major peak plus some minor peaks will be presented at the output. As shown in Figure 1.10(b), the major peaks are located at the overlapped peaks of these devices, showing a Vernier FSR of the least common multiple of total FSR values, and the height of major peaks is determined by the amount of overlap. When the RI above the sensing device changes, the major peak shifts ($\Delta\lambda_{\text{FSR}}$), i.e., $\Delta\lambda_{\text{max}} = m\Delta\lambda_{\text{FSR}}^{\text{ref}}$ [71]. In this way, the Vernier effect cascaded sensor system yields an ultra-high sensitivity which is given by [71]:

$$S = \left(\lambda_{\rm maj}/n_{\rm eff}\right) \left[\frac{\Delta \lambda_{\rm FSR}^{\rm ref}}{\left(\Delta \lambda_{\rm FSR}^{\rm ref} - \Delta \lambda_{\rm FSR}^{\rm sen}\right)}\right] = MS_0 \tag{1.15}$$

where λ_{maj} is the wavelength of the major peak, $\Delta\lambda_{FSR}^{ref}$ and $\Delta\lambda_{FSR}^{sen}$ are the FSRs of reference and sensing devices respectively, and S_0 is the actual sensitivity of the single sensing device. Thus, the sensitivity of the optical sensor based on Vernier effect cascaded devices is *M* times improved than that of a single device, without requiring a narrow linewidth tunable light source or a high-resolution readout system. The trade-off is that the readout is quantized, thus potentially limiting the minimum detection limits. Claes et al. developed cascaded MRRs with very large roundtrip lengths where FSRs difference is smaller than the FWHM of resonance peaks, resulting in the major peak's spectral period of [72]:

$$\Delta \lambda_{\rm FSR}^{\rm maj} = \frac{\Delta \lambda_{\rm FSR}^{\rm ref} \times \Delta \lambda_{\rm FSR}^{\rm sen}}{|\Delta \lambda_{\rm FSR}^{\rm ref} - \Delta \lambda_{\rm FSR}^{\rm sen}|}.$$
 (1.16)

By introducing a fitting procedure to reduce the smallest detectable wavelength shift, they obtained the Vernier effect-based sensing system which releases the limitation to the sensitivity by the $\Delta \lambda_{\text{FSR}}^{\text{ref}}$ [72].



Figure 1.10: (a) Illustration of the Vernier effect sensing system consisting of two cascaded MRRs with different FSRs. The sensing ring is exposed to RI changes in its environment, while the reference ring is covered by the cladding. (b) Illustrations of calculated transmission spectra of the reference device ($\Delta\lambda_{FSR1}$), sensing device ($\Delta\lambda_{FSR2}$), and cascaded system, respectively. Red-dashed lines represent transmission spectra after a RI change above the sensing device, showing an amplified wavelength shift in the cascaded system.

1.3.3 Sensitivities comparison

A sensor performance results comparison in the field of silicon photonic biosensors is presented in Table 1.2 along with different architectures as well as strategies to improve the S and DL values. Due to un-unified units of DL among different articles, bulk sensitivities in the unit of wavelength (or phase) shift per refractive index change are estimated from the results in the publications to serve as a comparison criterion. Moreover, other parameters and performance metrics such as light polarization and wavelength, system and intrinsic detection limits, and Q are also presented.

From the experimental results presented in Table 1.2, bulk sensitivities are enhanced in sensing configurations applied by performance-improving strategies. However, their detection limit values show no growth but a downward trend for slot and SWG waveguide-based sensors. That matches well with the recently published work by Kita et al., who found out that sensor performance of slot and SWG waveguides are not truly better than strip waveguide for sensing [95]. By proposing

Туре	Configuration	Strategy	Mode	Q	S _{bulk}	sDL	iDL
				(× 10 ³)	(\mathbf{RIU}^{-1})	(RIU)	(RIU)
Interferometer	MZI	Vernier	TE	N/A	$2.15 \times 10^4 \text{ nm}$	N/A	N/A [73]
		Suspended	TE	N/A	740 nm	N/A	4×10^{-5} [74]
		Slot	TE	N/A	$1730 imes 2\pi$ rad	1.29×10^{-5}	N/A [75]
		1.31 µm Wvl	TE	N/A	$540 imes 2\pi$ rad	N/A	N/A [76]
		N/A	TM	N/A	$460 imes 2\pi$ rad	$3.3 imes 10^{-5}$	N/A [16]
		N/A	TE	N/A	$300 \times 2\pi$ rad	N/A	N/A [58]
Microcavity	Ring	Vernier/suspended	TM	N/A	4.6×10^5 nm	N/A	4.8×10^{-6} [77]
		Vernier	TM	15	2.43×10^4 nm	N/A	N/A [78]
		Vernier	TE	20	1.3×10^3 nm	5.05×10^{-4}	N/A [79]
		Slot/critical coupling	TE	6	$1.3 \times 10^3 \text{ nm}$	N/A	$\leq 10^{-4}$ [80]
		Multi-box SWG	TE	2.6	580 nm	$1.8 imes10^{-5}$	1.02×10^{-3} [81]
		SOE-SWG	TE	1.49	575 nm	N/A	$1.8 imes 10^{-3}$ [82]
		SWG	TE	7	490 nm	2×10^{-6}	5.5×10^{-4} [83]
		SWG	TM	9.8	429 nm	N/A	3.71×10^{-4} [84]
		Slot	TE	0.33	298 nm	4.2×10^{-5}	1.59×10^{-2} [85]
		Suspended	TM	12	290 nm	N/A	N/A [86]
		Thin WG	TM	4.5	270 nm	N/A	1.2×10^{-3} [18]
		N/A	TM	10.1	200 nm	N/A	$7.5 imes 10^{-4}$ [18]
		Thin WG	TE	24	133 nm	N/A	5×10^{-4} [62]
		1.31 µm Wvl	TM	33.5	113 nm	N/A	1.49×10^{-3} [37]
		1.31 µm Wvl	TE	9.8	91 nm	N/A	3.5×10^{-4} [37]
		N/A	TE	15	38 nm	N/A	2.7×10^{-3} [37]
	Disk	N/A	TM	16	142 nm	N/A	6.8×10^{-4} [20]
		Suspended	TM	0.1	130 nm	$8 imes 10^{-4}$	1.18×10^{-1} [87]
		N/A	TE	33	26 nm	N/A	1.8×10^{-3} [20]
Photonic crystal	2D	Slot	TE	50	$1.5 \times 10^3 \text{ nm}$	$7.8 imes 10^{-6}$	2.07×10^{-5} [88]
		N/A	TE	0.4	200 nm	2×10^{-3}	1.88×10^{-2} [46]
		Ring-slot	TE	11.5	160 nm	N/A	8.75×10^{-5} [89]
	1D	Slot	TE	174	815 nm	N/A	1×10^{-5} [90]
		N/A	TE	3	130 nm	7×10^{-5}	4×10^{-3} [91]
Bragg grating	Phase-shifted	Multi-box SWG	TE	6.2	610 nm	N/A	4×10^{-4} [92]
		Slot	TE	15	340 nm	N/A	3×10^{-4} [93]
		1.31 µm Wvl	TM	76	106 nm	N/A	1.6×10^{-4} [37]
		N/A	TE	27.6	59 nm	N/A	$9.3 imes 10^{-4}$ [94]
	Uniform	N/A	TE	N/A	182 nm	N/A	N/A [21]

Table 1.2: Performance metrics comparison of selected sensors (WG = waveguide, Wvl = wavelength, 1.55 μ m where not specified). Our work is presented in bold.

a dimensionless figure of merit:

$$FOM = \frac{\gamma_{\text{clad}}}{\alpha_{\text{s}} \times \lambda} \tag{1.17}$$

where γ_{clad} is the waveguide mode sensitivity ($\gamma_{clad} = \frac{\partial n_{eff}}{\partial n_{clad}}$), and α_s is the scattering loss per unit length, both waveguide mode sensitivity and roughness scattering loss are taken into account for the comparison of various waveguide geometries by the authors. The model predicts that properly engineered TM-polarized strip waveguides claim the best performance compared to slot and SWG-based waveguides owing to their reduced propagation loss and longer accessible optical-path length [95].

1.4 Label-free detection

Generally, two approaches for optical detection are employed by most biosensors: label-based detection and label-free detection, as shown in Figure 1.11 below. In labelled detection, a label is defined as an additional molecule that is chemically or temporarily attached to the immobilized target to enhance the quantitative signal. Examples include, but are not limited to, a dye molecule (chromophore), a fluorescent tag, or an enzyme. This labelling process can achieve an ultra-low DL (on the order of sub-parts-per-trillion) and provide additional specificity via secondary amplifications [25]. However, it requires sophisticated reagent selection and pairing, in addition to reagent modification, including synthesis and purification, which potentially changes intrinsic properties of the capture probe and/or target molecules [96] and dramatically increases the cost and complexity of the assays. Moreover, due to the need for additional steps to perform label-based detection, it is ill-suited for real-time kinetic monitoring. To contrast, label-free detection has emerged as an appealing alternative to labelled detection, utilizing native molecular properties such as molecular weight (MW), RI, and molecular formal charge (FC) for target molecule monitoring. Label-free detection is not without its own drawbacks, as the method is only capable of providing sensitive and specific detection if non-specific binding (NSB) is low, or if the assay has sufficient controls to subtract the contribution of NSB. Additionally, label-free detection requires sufficient signal to be gen-



Adapted from: Flueckiger, Jonas. "Enhancing the performance of silicon photonics biosensors." PhD diss., University of British Columbia, 2017.

Figure 1.11: (a) Schematic of labelled bioassays, which involves a secondary amplification. Biomolecules, such as antibodies with a fluorescent tag, bind to the immobilized target analyte and provide signal amplification when stimulated. (b) Schematic of label-free biosensing. The bio-functionalized sensor continuously monitors any additional bound mass caused by captured target analytes.

erated upon binding for the sensor to differentiate signal from noise; this can limit label-free detection for certain applications with especially low molecular weight target species, or targets that do not readily interact with specific capture probes/-chemistries. Even with these limitations, a large number of biosensors designed for label-free detection have been investigated in the recent research literature [97–99], largely because the method greatly simplifies assays, can reduce both the time and number of steps required, and eliminates experimental uncertainty induced by the labelling process [100]. Additionally, label-free detection is highly amenable to the real-time kinetic evaluation of molecular binding and rapid quantification of analytes.

Since the first label-free optical biosensor was commercialized in 1990 by Biacore, Inc. [9], an entire field has arisen developing new platforms for label-free biosensing, driven largely by the appeal of addressing the unmet need in medical diagnostics, biosensing, and environmental/biohazard/threat monitoring. Among the new transducers, optical devices based on the SOI platform are among the most promising. Their highly compact footprint, allowing simultaneous multiplexed detection on a single chip, and low fabrication cost in high volumes with CMOS-compatible processes, make them cheap enough to be considered fully disposable. Table 1.3 gives an overview of a wide variety of exemplary target analytes, arranged in descending molecular weight, that have been detected using label-free SOI-based biosensors, as well as their reported DLs. This survey demonstrates that SOI-based optical biosensors have a wide detection range for analytes with MWs on the order of kilodalton (kDa). For large molecules like micrometer-sized cells and bacteria on the order of megadalton (MDa) or higher, their sizes may exceed the evanescent field range of the sensor and cause an invalid result. For small molecules (normally less than 500 Da), a detectable signal is difficult to achieve, especially for low concentrations, due to the low sensitivity or high noise level of SOI-based sensors.

1.5 Optical sensing system integration

To satisfy the need for system operations towards clinical and home healthcare diagnosis, integration is one of the key challenges to be solved [129]. The SOI platform is appealing since it offers the potential of optical component integration onto the same substrate. In recent years, a massive amount of effort has been made to integrate multiple functions to chip-scale silicon PICs, such as on-chip fluidic handling and optical analysis, as well as data processing [130]. These integrated sensing architectures show the ability for a high-density, lab-on-a-chip, and portable biosensing platform in the application of POC medical diagnosis. Here we review research directed towards the integration of microfluidics, lasers, sensing devices and photodetectors (PDs) on Si substrates for biosensing applications.

1.5.1 Optofluidic integration

Microfluidic systems have been regarded as an essential tool for modern biosensing research due to outstanding advantages such as low sample consumption, *in-situ* manipulation, short analysis time, controlled transportation, and high throughput [131, 132]. Recently, a synergy technique called optofluidics has emerged, which integrates microfluidics and photonic architectures to enhance each entity's function and performance [133]. Introducing optofluidics to silicon photonic biosensing systems not only combines fluid and light for improved sensing capability and simplification of microsystems but satisfies the function of on-chip, label-free, real-

Table 1.3: Overview of selected biomolecules that have been detected by optical sensors using label-free method (AIV = Avian influenza virus, HSV= Herpes simplex virus, BPMV = Bean pod mottle virus, HPV = Human papillomavirus, SA = Streptavidin, HSA = Human serum albumins, PSA= Prostate specific antigen, CRP = C-reactive protein, GM = Gentamicin, BPT = biphenyl-4-thiol, CFU = colony-forming unit, HAU = hemagglutination unit, VP = viral particle).

Analyte	Target	Weight	Туре	Material	DL
Cell	E. coli	1 pg	MRR	Hydex	10 ⁵ CFU/mL [101]
			MRR	Si	10 ⁸ CFU/mL[102]
Virus	AIV	542 MDa	MZI	Si_3N_4	$5 imes 10^{-4}$ HAU/mL [103]
	HSV	96 MDa	YI	Si_3N_4	850 VP/mL [104]
	BPMV	7 MDa	MRR	Si	1.43 pM [27]
	HPV	5 MDa	PhC	Si	1.4 nM [105]
Protein	IgG	150 kDa	PHC	Si 1 ng/mm ² [10	
			MZI	Polymer	3.1 nM [76]
			Vernier MRR	Si	47.3 nM [107]
	avidin/SA	55-68 kDa	MZI	SiO _x N _y	2.14π/nm [108]
			PhC	Si	2.5 fg [24]
			PhC	Si	344 pm/nm [109]
			Slot MZI	Si_3N_4	18 fM [110]
			PhC	Si	49 fM [111]
			MRR	Si	60 fM [112]
			MRR-MZI	Si	20 pM [113]
			MRR	SiO ₂ /Si _x N _y	0.1 nM [114]
			MRR	Si	0.15 nM [17]
			Slot disk	SiN _x	0.55 nM [115]
	HSA	67 kDa	YI	Si ₃ N ₄	20 fg/mm^2 [116]
			MRR	Si	3.4 pg/mm ² [117]
	PSA	28 kDa	MRR	Si	0.4 nM [118]
			Slot MRR	SiN	1.79 nM [119]
	CRP	25 kDa	MZI	Si _x N _y	84 fM [120]
			MRR	Si	0.4 nM [121]
			MZI	SiN	0.78 nM [122]
Nucleic acid	RNA	7-40 kDa	MRR	Si	53 fM [123]
			MRR	Si	150 fM [29]
			Slot MZI	Si_3N_4	1 nM [124]
	DNA	7-12 kDa	MZI	Si_3N_4	300 pM [125]
			Slot MZI	Si_3N_4	1 nM [110]
			MRR	Si	1.95 nM [28]
			PHC	Si	19.8 nM [126]
			MRR	Hydex	100 nM [101]
Small molecule	GM	478 Da	PhC	Si	0.1 nM [127]
	BPT	186 Da	PhC	Si ₃ N ₄	N/A [128]

time detections. In addition, optofluidic sensors are extremely suitable for evanescent field RI detection, since the change of RI scales with the analyte bulk concentration or surface density, rather than the number of molecules in total [133].

Polydimethylsiloxane (PDMS) has become the most popular material in the academic microfluidics community since it is inexpensive, easy to fabricate, flexible, optically transparent, and biocompatible [134]. More importantly, PDMS material can be permanently bound to SiO_2 substrates after oxygen plasma treatment [135], which provides a simple and fast approach to build leakage-free microfluidic channels on SOI-based sensors. Many silicon photonic devices including MZIs [113, 136], MRRs [137] and PHCs [91, 127, 132] have employed PDMS microfluidic systems mounted on top as a convenient optofluidic delivery method for analyte detection. However, PDMS also shows some drawbacks. On the one hand, PDMS is not suitable for the integration or deposition of electrodes directly on the surface, and has problems such as adsorption of small hydrophobic molecules, swelling in organic solvents, water permeability, and incompatibility under very high-pressure operations [138]. On the other hand, due to the irreversible bonding process, chips are not reusable after mounting the PDMS microfluidic block, and most of the area on the chip only serves as a mechanical support for the fluidic inlet and outlet but not for sensing, which negatively impacts the unit cost.

Another commercially available material, negative tone photoresist SU-8, has been employed for on-chip optofluidics recently. SU-8 was originally developed as a high-resolution photoresist for the microelectronics industry. Because of its transparency in the near-infrared spectrum and biocompatibility, a thin layer of SU-8 coating with microfluidic patterns has been investigated on silicon photonic biosensing systems [139], which improves the alignment precision compared to PDMS microfluidics bonding. Furthermore, SU-8 can also be used as a cover material for interface passivation of on-chip electrical connections due to its highresolution patterning and insulation abilities. However, the manufacturing process of the SU-8 microfluid requires the use of cleanroom facility equipment involving complex and numerous processing steps, which hinders mass production at a low price. In addition, variation in conditions such as humidity and SU-8 composition may affect fabrication protocols, contributing to batch-to-batch variability [140]. Other materials such as glass [141], polycarbonate (PC) [142], cyclic olefin copolymer (COC) [143] and epoxy [144] were also reported for the on-chip optofluidic integration.

1.5.2 Optoelectronic integration

One of the biggest roadblocks towards the large-scale commercialization of photonic biosensors is the low-cost high-yield integration of light sources to operate reliably while consuming minimal power. These goals are usually traded-off against each other with the choice of platform for integrating the light source, the sensor device, and the photodetector (PD) to achieve a complete lab-on-a-chip system. For instance, to benefit from a high-yield and low-cost production, leveraging existing CMOS fabs seems to be the ideal solution. This requires the integration of these three elements on a single Si CMOS-compatible die. However, integrating the active laser source with the passive sensor device and the PD remains a challenge. Several techniques utilized for the chip-scale optoelectronic integration are presented below, and advantages brought as well as challenges faced by each method are highlighted.

On-chip lasers

Driven by the promises lasers on Si hold for optical communication [145], several groups across the world have demonstrated integrated lasers on Si dies implemented using either group IV materials (Si or Ge) or group III/V compounds [146]. While using group IV elements seem to be an appealing and practical solution in terms of cost and portability, existing methods using Si cannot yet render an electrical input/output (I/O) based lab-on-a-chip because they rely on optical pumping mechanisms [147, 148], making it an unattractive solution at the moment. Electrically-pumped Ge lasers integrated on Si, however, have been demonstrated [149]. Despite its indirect bandgap, straining and n-doping Ge can tailor its bandgap to make it direct [150]. Repercussions of this approach are high threshold currents [149], thus increasing the total power budget of the biosensors.

On the other hand, III/V lasers integrated on Si have been demonstrated with much higher efficiency in comparison to Ge, thanks to their direct bandgap and superior gain characteristics. While monolithic integration of III/V compounds on Si seems to be the optimum solution for ease of portability and highest density integration, the biggest bottleneck towards the direct monolithic growth of III/V compounds on Si lies in the lattice and thermal expansion coefficient mismatch between the Si material and III/V compounds [146]. To solve this problem, three main approaches have been demonstrated to integrate III/V lasers on Si chips: (1) direct mounting, (2) hybrid approaches through direct and indirect bonding heterogeneous integration, and (3) monolithic integration using sophisticated growth techniques.

On-chip detectors

For a lab-on-a-chip system with electrical I/Os, an on-chip photodetector is required to convert the light signal for further processing. There are several on-Si PDs implemented either using III/V compounds or using group IV elements such as Si or Ge. The choice of PDs depends on the detection wavelength of interest. Wang et al. have heterogeneously integrated III/V PDs on Si substrate for operation at a wavelength of [151]. Other techniques explored include thermo-electric PDs [151, 152]. However, across the C-band, besides III/V compounds [153], Ge and Si could be used for photodetection. The main advantage of using Si or Ge is their ease of fabrication with a CMOS fab. Despite the transparency of silicon at the Cband, doping Si can increase the Siwaveguide sensitivity to incoming light across the C-band either due to surface states [154], or due to the introduction of midband-gap defect states [154-157]. Si-based defect-mediated PDs, however, suffer from either low responsivities or large photoconductive gain at the expense of a much larger dark current [156], which is undesirable for biosensing applications. Ge-based PDs, however, have superior characteristics. Recent results showed Ge on Si PDs with a high responsivity of 0.74 A/W and low dark currents of less than 4 nA [158]. These characteristics make Ge-based PDs ideal for biosensing at a wavelength of 1.3 or 1.5 µm in the SOI platform.

1.5.3 Readout and noise

For conventional evanescent field biosensing techniques, two aforementioned methods are usually employed for the quantitative detection of analytes at the sensor surface in real-time: the first one is monitoring the wavelength (or phase) shift in the transmission spectrum through scanning the input light source wavelength, which allows a large dynamic range for sensors; the other one is detecting the transmission intensity change caused by shifts at a fixed wavelength and providing precise detection with a very small concentration of analytes [159, 160]. Both of these spectral-domain approaches require precise optical spectrum scanning or processing systems, such as a wavelength-tunable laser, high-resolution photodetector or optical spectrum analyzer. Correspondingly, two types of spectral noise sources, wavelength noise and intensity noise, are categorized: wavelength noise $(\sigma_{wavelength})$ is mainly generated from the light source wavelength shift and thermally influenced fluctuations of the sensor; whereas intensity noise ($\sigma_{intensity}$) is caused by light source intensity fluctuations, the variation of input coupling, and the PD noise [161]. Another important factor, the spectral resolution ($\sigma_{\text{resolution}}$) of the system setup, can also limit the precision of the spectral location, which highly depends on the measurement setup, i.e., the laser or the optical readout. Therefore, the total noise variance in the sensing system can be approximated by summing all the individual noise variances [43]:

$$3\sigma = 3\sqrt{\sigma_{\text{wavelength}}^2 + \sigma_{\text{intensity}}^2 + \sigma_{\text{resolution}}^2}.$$
 (1.18)

Several approaches can be applied to improve the system noise for silicon photonic sensors. As mentioned before, Q plays an important role in determining the DL of a resonant sensor. That is because having a high Q (narrow FWHM) can filter the spectral noise effectively and lead to a low spectral deviation from the actual extremum [43]. Another one is introducing optical spectrum curve fitting, which is a powerful tool to enhance the spectral resolution. Taking into account of the entire spectrum, a fitting process can improve the eventual signal-to-noise ratio (SNR) by \sqrt{N} , where N is the total number of data points in the spectrum of interest [36, 161]. By applying this peak fitting algorithm to resonant sensors, a wavelength measurement precision much smaller than both the light source linewidth and the peak FWHM is achieved [161], with a factor of approximately 10 to 10^3 [45]. Therefore, the system DL with an improved linewidth in the spectrum readout can greatly enhance sensor performance as compared to the intrinsic DL using the peak linewidth according to Equation 1.5 and 1.10. However, due to the approximate fitting, many factors can lead to resonant peak shape deviations from the ideal function, such as peak asymmetry, broadening, splitting, and even an excessive Q [161]. In term of interferometer-based sensors, the spectral shape is simply a squared cosine, which can be directly interpreted as the fit function, offering a fit curve as accurate as possible [162]. Thermally influenced fluctuation is the noise source for silicon photonic sensors which cannot be declined by Qs and curve fitting. It generates from the peripheral temperature variation of the sensor. Besides, different materials in the sensor (cladding, waveguide core, and BOX) respond differently to the variation, i.e., different thermo-optic coefficients (TOC = $\Delta n/\Delta T$, where ΔT is the temperature change). Therefore, approaches such as introducing a reference device with identical TOC characteristics, or designing thermal-insensitive (athermal) architectures are commonly employed to reduce the noise caused by temperature fluctuations [161].

1.5.4 State-of-the-art CMOS-chip packaging

Compared to traditional benchtop sensors and instrumentation, biosensors that rely on CMOS processes offer lower cost, lower power and smaller size with a highdensity on-chip sensing array [163]. In terms of lab-on-a-chip monitoring, the primary challenge is the integration of sensing arrays interfaced with fluid samples and electrical interconnects for data processing on CMOS substrates. Furthermore, die-level CMOS substrates are always millimeter-sized which obstructs the on-chip microfluidics and electrical interconnections integration for high-throughput.

To overcome these difficulties, several post-CMOS approaches have been investigated as system-level packaging to implement electronic and biological detection functions. Fluid barrier materials, such as PDMS, epoxy, SU-8, oxide/nitride, and parylene, have been employed for integrating CMOS chips with microfluidics. Li et al. reported a chip-in-package process utilizing wire bonding technology for the die-level on-CMOS biosensor integration [164]. By depositing a 2- μ m-thick parylene layer as the insulating coating, the biosensor is enabled for operations in liquid with a good functionality of CMOS electronics [164]. Huang et al. developed a lab-on-CMOS platform for electrochemical microsystems by using ox-

ide/nitride/oxide (ONO) passivation layers, which allows the functional integrity of multi-channel microfluidic structures and on-CMOS electrodes [165]. For the size disparity between the CMOS chip and on-chip microfluidics, die-level CMOS chips have been encapsulated into a substrate carrier which enlarges the surface area for further processes.

In 2014, Datta-Chaudhuri et al. presented a simple packaging method for dielevel CMOS foundry-fabricated chips, which are embedded in epoxy handle wafer for a level, enlarged surface, allowing subsequent post-processing and microfluidic integration [166]. Parylene-C was selectively exposed to the surface for the passivation of electrical connections. Due to the flat surface around the chip, good electrical continuity of fan-out metal traces from the chip to the edge of the wafer is achieved, enabling the subsequent off-chip data communication [166]. Similarly, a CMOS-compatible epoxy chip-in-carrier process was developed by Lin et al. [167]. By introducing a planar screen-printed silver ink metallization technique with mounted multi-channel PDMS microfluidics on the device's surface, electrochemical and microfluidic experiments were evaluated by interconnect resistance measurements, showing high effectiveness for lab-on-CMOS applications to achieve desired capability with high yield and low material and tool cost [167].

1.6 Research objectives

The main contribution of the work presented in this dissertation is giving the potential to pave the path of the development of low-cost, high-sensitive, disposable optical biosensors. Compared to SPR techniques, silicon photonics is a good platform in the development of lab-on-a-chip sensing instrument with low-cost, leveraging the CMOS foundries around the world. However, due to the strong optical confinement of the standard strip and rib waveguides, the sensitivity of the silicon photonic-based sensors is not satisfiable for real applications. Moreover, optical setups working in the near-infrared range are expensive and not easy to implement. Other optical components, such as tunable lasers and optical spectrum analyzers are rather high-cost, which restricts the application of silicon photonic-based sensors, There is still a gap between the application of silicon photon sensors and the commercialization of SPR techniques. Therefore, in this dissertation, we focus on the development of a silicon photonic-based biosensing system to enhance sensor performance and reduce the overall unit cost, in order to approach the sensitivity competitive to the SPR system but with smaller set-up size and reduced cost.

The objectives of this dissertation research can be divided into two parts. The first one is to develop novel sensing architectures with greater sensor performance than other competitive silicon photonic devices, which is accomplished by developing and characterizing new sensing configurations, such as multi-box SWG waveguide-based MRRs and PSBGs, and substrate overetched SWG waveguide-based MRRs. Another one is to design cost-effective, foundry-compatible silicon photonic sensing systems, which is achieved by using low-cost broadband light sources and on-chip photonic readout leveraging the CMOS-compatible foundries, and introducing the lab-scale Fan-Out Wafer-Level Packaging (FOWLP) process.

1.7 Dissertation organization

The remainder of this dissertation is organized into four main chapters as follows:

Chapter 2 describes a sensitivity improvement strategy for the sub-wavelength waveguide-based microring resonator sensor by using the anisotropic substrate overetch process. The chapter also includes the design and simulation approaches and the comparison of theoretical models with experimental results. Details about the etching process and sensor performance characterization are presented. Moreover, we also mention detailed materials and methods used for optical and biosensing experiments throughout this dissertation in this chapter.

Chapter 3 describes the development of the novel multi-box sub-wavelength metamaterial by applying periodic sub-wavelength configurations in both propagation and transverse directions. Details about the design, layout, simulations, and measurements are also given in this chapter. Sensor performance of the proposed waveguide is characterized through microring resonator and phase-shifted Bragg grating architectures. To further demonstrate the capability in the utilization of monitoring small molecule interactions, a biotin-avidin affinity model is employed to the proposed sensor.

Chapter 4 describes two novel and appealing approaches to develop a cost-

effective sensing system in the SOI platform. The first one is using two cascaded microring resonators, one of the rings is an in-resonator photoconductive heaterdetector, which works as the on-chip readout component. The other one is using a phase-shifted Bragg grating as arms for the Mach-Zehnder interferometer sensing architecture and an intensity interrogation scheme. Design and simulation details, as well as sensing characterizations, are included in the chapter.

Chapter 5 describes the development of the lab-scale Fan-Out Wafer-Level-Packaging process, which enables a low-cost approach to integrate the sensing device, microfluidics, and optoelectronic readout onto the same substrate. The chapter also shows the measurement setup building, and sensing characterizations of the packaged silicon bio-chip on this setup.

Chapter 6 describes the main conclusions and future research direction.

Chapter 2

Substrate-overetch Micro-ring Resonator Sensors

2.1 Introduction

Resonators are devices or systems that present resonance or resonant behaviours, which oscillate with larger amplitude at resonant frequencies, than at other frequencies. The discovery of the optical resonator can be traced back to 1899, which were first proposed by two French physicists Fabry and Pérot when they constructed a parallel-plate resonator as a multi-pass interferometer. Microring resonators (MRRs), also known as ring resonators or racetrack resonators, consist of an optical waveguide which is looped back on itself, with some coupling to the outside [53]. Based on the number of the directional coupler as coupling, there are two MRR configurations, i.e., all-pass and add-drop ring resonators. Integrated MRRs play an important role in the success of silicon photonics, thanks to the unprecedented small size that silicon enables [13]. In the past decades, silicon photonic integrated MRRs have provided attractive solutions for applications in communications [168, 169], signal processing [170], quantum computing [171], as well as biosensing [18].

A theoretical understanding and derivation of fundamental principles will be introduced for the optical waveguide, and the ring resonator configuration in Section 2.2, which is key for the sensor architecture optimization and performance comparison. In Section 2.3, a novel sub-wavelength grating (SWG)-based sensing configuration is investigated for sensor performance enhancement. By introducing a substrate-overetch (SOE) geometry, the proposed SOE-SWG waveguide shows enhanced light-matter interaction and a reduced group index. Details about the proposed SOE-SWG waveguide-based MRR sensor modelling and simulations are also presented. Furthermore, sensor performance, including bulk and surface sensitivities, are characterized for the proposed architecture, and compared with conventional SWG-based MRRs.

2.2 Principles of microring resonators

2.2.1 Optical waveguides

The heart of any optical biosensor is an optical waveguide, which was originally developed for telecommunication applications. The mechanical stability, flexible geometry, noise immunity and efficient light-conducting over long distances make the optical waveguide also well-suitable for implementation in sensor applications [9]. The strip waveguide, also known as ridge waveguide, is one of the most common silicon photonic waveguides with a scale size on the order of sub-microns (Figure 2.1), which is widely used for routing as it offers strong optical confinement within the waveguide, thus a tight bend radii [53]. There is a buried oxide layer (BOX) underneath the top thin silicon layer, which can isolate the waveguide from the bottom silicon substrate to avoid the substrate optical leakage. As shown in Figure 2.1, the thickness of the silicon layer is 220 nm, with the BOX layer of 2 μ m and the Si substrate of ~750 μ m. This configuration is widely used for silicon photonics, as well as in CMOS foundries. We use this SOI wafer throughout this dissertation.

As we know, light is an electromagnetic wave. Therefore, Maxwell's equations can perfectly describe the propagation of light inside the waveguide:

$$\nabla \times \vec{E} = -\frac{\partial \vec{B}}{\partial t},\tag{2.1}$$


Figure 2.1: Schematic of a strip waveguide cross section with a rectangular shape (not to scale).

$$\nabla \times \vec{H} = -\frac{\partial \vec{D}}{\partial t} + \vec{J}, \qquad (2.2)$$

and

$$\nabla \cdot \vec{D} = \rho, \qquad (2.3)$$

$$\nabla \cdot \vec{B} = 0, \tag{2.4}$$

where \vec{E} and \vec{H} represent electric and magnetic field intensities, respectively, \vec{D} is the electric displacement field, \vec{B} is the magnetic flux density, \vec{J} is the electric current density ($\vec{J} = 0$ in optical waveguides), and ρ is the charge density. For the light propagating in the Z direction (in Figure 2.1), the modal fields can be written in separable form:

$$\vec{E}(\vec{r},t) = Re[\vec{E}(x,y)e^{j(\omega t - \beta z)}], \qquad (2.5)$$

$$\vec{H}(\vec{r},t) = Re[\vec{H}(x,y)e^{j(\omega t - \beta z)}], \qquad (2.6)$$

where β is the propagation constant along the Z direction. Waveguide modes can be regarded as transverse resonances of the field in an optical waveguide. The confinement factor (Γ) is defined as the portion of the light guided in the waveguide,

which is the fraction of the total power residing in the waveguide core [172–174]:

$$\Gamma = \frac{\frac{1}{2} \iint_{\text{core}} Re(\vec{E} \times \vec{H}^*) \cdot \hat{z} \cdot dx dy}{\frac{1}{2} \iint_{\text{total}} Re(\vec{E} \times \vec{H}^*) \cdot \hat{z} \cdot dx dy} = \frac{\frac{1}{2} n_{\text{core}} c_0 \varepsilon_0 \iint_{\text{core}} |E|^2 \cdot dx dy}{\frac{1}{2} \iint_{\text{total}} Re(\vec{E} \times \vec{H}^*) \cdot \hat{z} \cdot dx dy}.$$
(2.7)

The confinement factor is related to the mode number, the fundamental mode (or the lowest mode number) has the highest confinement (Γ close to 1). For higherorder modes, mode confinement is decreased until the waveguide no longer supports the proposed mode [25]. The confinement factor depends on the refractive index contrast between the waveguide core and the surrounding media. In the aspect of sensing, we care about the fraction of power outside of the waveguide (1- Γ) more.

The refractive index (RI) is a material-based property, which describes the increase in the phase change per unit length due to the medium. If we assume the magnetic permeability is the same as free space ($\mu_r \approx 1$), the refractive index can be defined as:

$$RI = \sqrt{\varepsilon_{\rm r}(f)},$$
 (2.8)

where ε_r is the relative permittivity of the material and *f* is the frequency of light. As for the effective refractive index, it has a similar meaning as the RI, but it is not just a material property, which includes the RI contributions from all the materials experienced by the optical mode as well as the waveguide geometry itself. The effective index can also be regarded as the ratio of the phase velocity in vacuum (*c*) to the phase velocity in the waveguide ($v_{p,m}$) [25]:

$$n_{\rm eff,p,m} = \frac{c}{v_{\rm z,p,m}} = \frac{k_{\rm z,p,m}}{k_0} = \frac{\beta_{\rm p,m}}{k_0},$$
 (2.9)

where $\beta_{p,m}$ is the mode propagation constant ($k_{z,p,m} \equiv \beta_{p,m}$) and *k* represents the wavenumber.

As we know, the effective index of the mode propagation in the waveguide is not only related to the geometry but also to frequency (f) of the light. The dispersion of the waveguide describes the effective index change as a function of the frequency $(n_{\text{eff}}(f))$. This means that the phase velocity (i.e., the group velocity) is changing with frequency change. There are two contributing factors to the dispersion of a waveguide mode. The first one is the material dispersion, which consists of all refractive index dispersions of given materials. The other one is the waveguide dispersion, which describes the frequency dependence of the waveguide mode by the 2D confinement.

The group velocity refractive index, usually called the group index (n_g) , is defined as:

$$n_{\rm g}(f) = \frac{c}{v_{\rm g}(f)} = n_{\rm eff}(f) - \lambda_0 \frac{\partial n_{\rm eff}}{\partial \lambda}, \qquad (2.10)$$

where v_g is the group velocity describing the speed of the pulse envelope in the dispersive medium. It can be seen that the group index depends on the wavelength, similar to the effective index.

The power travelling outside of the waveguide is called the evanescent field (as discussed in Section 1.2), which decays exponentially away from the waveguide surface [175]. As shown in Figure 2.2 below, most of the intensity is condensed within the waveguide core and decays to the surrounding media. For a waveguide slab in Figure 2.2(a) (confinement only in the y direction), the electric field decays near the surface can be expressed as [25]:

$$E(y) = E_0 e^{-y\frac{2\pi}{\lambda}\sqrt{n_{\rm eff}^2 - n_{\rm clad}^2}},$$
 (2.11)

where E_0 represents the electric field at the waveguide surface. While, for a waveguide core in Figure 2.2(b) (confinement in the x and y directions), the evanescent field decays at the center of the waveguide (E(x = W/2, y) and E(x, y = t/2)) [176]. Therefore sensor performance of an evanescent field-based sensor scales with the modal overlap of the electric field with the analyte: a higher surface sensitivity can be achieved close to the waveguide surface and will decrease as the increase of the adlayer thickness. If the refractive index perturbation factor (analyte) exceeds the decay length, or called the penetration depth (d), the next variation of the analyte will no longer change the propagation properties of the optical mode propagating in the waveguide. The decay length is usually referred to as the distance where the optical field intensity has decayed to 37% (1/e):

$$d_{1/e} = \frac{\lambda}{2\pi} \frac{1}{\sqrt{n_{\rm eff}^2 - n_{\rm clad}^2}}.$$
 (2.12)



Figure 2.2: Cross-sectional plots of the evanescent field intensity (**a**) of a waveguide slab in the y-z plane, and (**b**) a waveguide core in the x-y plane. The top surface of the waveguide is exposed to the analyte solution. *W* is the waveguide width, and *t* is the waveguide thickness.

It has been proved that the decay length for a strip waveguide in the quasi-TE mode is $d_{\text{TE}} = 121$ nm, and in the quasi-TM mode is $d_{\text{TM}} = 203$ nm, respectively [25].

There are two common choices for patterning the silicon devices: electronbeam lithography (EBL) and deep ultraviolet (DUV) lithography. EBL is performed by drawing the design directly onto the photoresist via an electron beam. This approach can achieve smaller feature sizes (~ 60 nm) with fast turn-around times, but at the cost of low throughput. DUV lithography is performed with a stepper that prints a copy of the mask design on to each segment of a larger wafer with high throughput. In addition, DUV offers complex systems to be reliability built with comprehensive photonic functionalities. Details about EBL and DUV process can be found in Ref. [177]. In this dissertation, all sub-wavelength devices (presented in Chapter 2 and Chapter 3) were fabricated through EBL due to the small sub-wavelength feature sizes. While, in Chapter 5, the integrated sensing system was realized by DUV, since it requires on-chip active components. However, it is worth to notice that the fabrication of both processes is not perfect, which means that the real silicon waveguide cross-section is not a perfect rectangle, and the sidewall is not 100% smooth. This is caused by the erosion of mask during etching and incomplete fully anisotropic silicon etching. Figure 2.3 shows the scanning electron microscope (SEM) image of the fabricated strip waveguide at ANT. From



Figure 2.3: (a) SEM image of a conventional strip waveguide fabricated at Applied Nanotools, with a designed width of 500 nm. The sample is tilted. (b) Cross-sectional SEM image of the aforementioned waveguide, presenting a trapezoidal-shape cross section.

Figure 2.3(b), we can see that the cross-section is not rectangular, with a small angle around 10° , which causes the upper and lower widths 437 and 522 nm, respectively. This geometric imperfection will change the waveguide mode profile and the effective index, thus affecting the performance of the sensors slightly.

2.2.2 Ring resonators

Resonant condition (constructive interference) describes the phenomenon when the optical length of the resonator is equal to a multiple of the wavelengths, as defined in Equation 1.8. The coupling is one of the fundamental components of a microring resonator, which consists of two waveguides as a 2×2 coupling region, as presented in Figure 2.4. Within the coupling, the input and output electric fields can be expressed as the coupling matrix:

$$\begin{bmatrix} E_1'(\boldsymbol{\omega}) \\ E_2'(\boldsymbol{\omega}) \end{bmatrix} = \begin{bmatrix} t & -i\kappa \\ -i\kappa & t \end{bmatrix} \begin{bmatrix} E_1(\boldsymbol{\omega}) \\ E_2(\boldsymbol{\omega}) \end{bmatrix}, \qquad (2.13)$$

where E_1 , E_2 and E'_1 , E'_2 are electric fields at input and output, for waveguide 1 and waveguide 2, respectively, *t* and κ represent the field straight-through transmission coefficient and cross-over coupling coefficient, (as presented in Figure 2.4), and ω is the angular frequency of the wave. The fraction of the power in the coupled (cross) and original (through) waveguides can be expressed as:

$$|\kappa|^2 = \frac{P_{\text{cross}}}{P_0} = \sin^2 \left(C \times L\right),\tag{2.14}$$

$$|t|^{2} = \frac{P_{\text{through}}}{P_{0}} = \cos^{2}\left(C \times L\right), \qquad (2.15)$$

where P_0 is the input power, P_{cross} and P_{through} are the power remaining in the through waveguide and coupled across the direction coupler, *L* is the length of the coupler, and *C* is the coupling coefficient. For a loss-less coupling region, $t^2 + \kappa^2 = 1$. *C* can be written as:

$$C = \frac{\pi \Delta n}{\lambda},\tag{2.16}$$

where Δn is the index difference of coupled waveguides between two eigenmodes. Another important parameter in the coupler is the coupling length (L_c), which is defined as the distance for all the power localized in the coupled waveguide. It depends on the wavelength (λ) and the distance (g) between the waveguides, and is given by:

$$L_{\rm c}(\lambda,g) = \frac{\lambda}{2 \times \Delta n(\lambda,g)} = \frac{\pi}{2 \times C}.$$
(2.17)

In the aspect of the ring resonator, the waveguides in the coupling region are connected to bend waveguides that also contribute to the coupling. Such as L = 0, light can be still coupled due to the waveguide bending [44]. Therefore, Equation 2.14 and 2.15 can be extended to include the contribution of the bend region:

$$|\kappa|^2 = \sin^2\left(\frac{\pi}{2L_c}[L+z_{\text{bend}}]\right),\tag{2.18}$$

$$|t|^2 = \cos^2\left(\frac{\pi}{2L_c}[L+z_{\text{bend}}]\right),$$
 (2.19)

where z_{bend} is the effective extra coupler distance from the non-parallel waveguides.

In an all-pass microring resonator (shown in Figure 2.5), only one 2×2 directional coupler is employed to feed the output back into its input. The relationship



Figure 2.4: Schematic of a standard 2×2 direction coupling region.

between E_{R1} and E_{R2} in the resonator is:

$$E_{\rm R1} = e^{-\alpha_0 L/2} e^{i\beta L} E_{\rm R2} \equiv a e^{i\phi} E_{\rm R2}$$
(2.20)

where *a* is the amplitude transmittance $(a = e^{-\alpha_0 L/2})$, and ϕ is the phase change $(\phi = (2\pi n_{\text{eff}}L)/\lambda)$, after each roundtrip, respectively, α_0 is the loss in dB unit per unit length, and *L* is the circumference of the resonator. According to Equation 2.13 and 2.20, we can obtain:

$$E_{\rm R2} = \frac{-i\kappa}{1 - tae^{i\phi}} E_{\rm in} \tag{2.21}$$

$$E_{\rm t} = \frac{t - ae^{i\phi}}{1 - tae^{i\phi}} E_{\rm in} \tag{2.22}$$

Therefore, if we set the input intensity is I_{in} ($I_{in} = |E_{in}|^2$), the intensity in the resonator (I_{Rq}), and the intensity at the through port (I_t) can be expressed as:

$$\frac{I_{\rm R2}}{I_{\rm in}} = \left|\frac{E_{\rm R2}}{E_{\rm in}}\right|^2 = \frac{\kappa^2}{1 + a^2 t^2 - 2ta\cos\phi}$$
(2.23)

$$T = \frac{I_{\rm t}}{I_{\rm in}} = \left|\frac{E_{\rm t}}{E_{\rm in}}\right|^2 = \frac{t^2 + a^2 - 2ta\cos\phi}{1 + a^2t^2 - 2ta\cos\phi}$$
(2.24)

where T is the intensity transmission at the through port.

For an add-drop microring resonator, the ring is coupled to two waveguides, the incident field is partly transmitted to the drop port, as shown in Figure 2.6. Different from the all-pass MRR, *a* turns to be t_2a in the add-drop MRR due to the second coupling. If no power is introduced from the add port ($E_{add} = 0$), E_t , E_d , and E_{R2} are defined as:



Figure 2.5: Schematic of a standard all-pass microring resonator.

$$E_{\rm t} = \frac{t_1 - t_2 a e^{i\phi}}{1 - t_1 t_2 a e^{i\phi}} E_{\rm in}$$
(2.25)

$$E_{\rm R2} = \frac{i\kappa_1}{1 - t_1 t_2 a e^{i\phi}} E_{\rm in} \tag{2.26}$$

$$E_{\rm d} = \frac{\kappa_1 \kappa_2 \sqrt{a} e^{-i\phi/2}}{1 - t_1 t_2 a e^{i\phi}} E_{\rm in}.$$
 (2.27)

The intensity transmissions at the through (T_t) and drop ports (D_d) are:

$$T_{\rm t} = \frac{I_{\rm t}}{I_{\rm in}} = \frac{t_1^2 + t_2^2 a^2 - 2t_1 t_2 a \cos \phi}{1 + t_1^2 t_2^2 a^2 - 2t_1 t_2 a \cos \phi}$$
(2.28)

$$D_{\rm d} = \frac{I_{\rm d}}{I_{\rm in}} = \frac{\kappa_1^2 \kappa_2^2 a}{1 + t_1^2 t_2^2 a^2 - 2t_1 t_2 a \cos \phi}.$$
 (2.29)

Several important parameters are of interest in designing microring resonators, including the extinction ratio (ER), free spectral range (FSR), quality factor (Q), full width at half maximum (FWHM, or called 3-dB bandwidth), as well as finesse (F).

The extinction ratio is defined as the ratio of the maximum to the minimum powers at the through or drop ports transmission in a resonator. In telecommunication applications, the ER is dictated by the sensitivity of the receiver for a certain bit error rate (BER). In the aspect of biosensing, a larger ER allows a higher signal-



Figure 2.6: Schematic of a standard add-drop microring resonator.

to-noise ratio, thus a robust resonant peak for wavelength shift tracking. The ER for the through (ER_t) and drop (ER_d) transmissions can be written as:

$$ER_{t} = 10\log_{10}\left[\frac{T_{\text{max}}}{T_{\text{min}}}\right] = 10\log_{10}\left[\frac{(t_{1}+t_{2}a)(t_{1}t_{2}a-1)}{(t_{1}-t_{2}a)(t_{1}t_{2}a+1)}\right]^{2},$$
 (2.30)

$$ER_{\rm d} = 10\log_{10}\left[\frac{D_{\rm max}}{D_{\rm min}}\right] = 10\log_{10}\left(\frac{1+t_1t_2a}{1-t_1t_2a}\right)^2.$$
 (2.31)

The case where the transmission of the resonator completely drops to zero is referred to as critical coupling. This situation happens when the coupled power (κ^2) equals the total round-trip loss inside the resonator (1 – a^2), in other words, when t = a ($t_1 = t_2 a$ for add-drop rings). The resonator shows the maximum ER when it is critical coupled.

The FSR, Q, and FWHM have been described in Chapter 1. The free spectral range describes the spacing between two adjacent resonances in the transmission of a resonator, which can be expressed in phase, wavelength or frequency. Due to the material and waveguide dispersion, the FSR value is not straightforward but wavelength dependent (Equation 1.12). The quality factor is a measure of the sharpness of the resonance relative to its central frequency (Equation 1.9). The full

width at half maximum or the 3-dB bandwidth is the bandwidth of the resonant peak at half of its maximum power, i.e. 3-dB decay of the original power in the logarithmic scale.

Finesse of a resonator is defined as the ratio of FSR and FWHM:

$$F = \frac{\Delta\lambda_{\rm FSR}}{\Delta\lambda_{\rm FWHM}} = \frac{\pi \times \sqrt{ta}}{1 - ta}.$$
(2.32)

It is a measure of the sharpness of the resonances relative to their spacing. F increases with the increase of a, and the decrease of κ .

2.3 Substrate-overetch resonator-based sensors¹

As mentioned in Section 1.3.2, sub-wavelength grating-based geometry has been considered to be an appealing metamaterial for the development of next-generation PICs. Due to its unique optical properties, SWG waveguide-based sensors have achieved a lot of studies, indicating enhanced sensor performance [83, 178]. Recently, Wangüemert-Pérez et al. proved that a narrower and thicker SWG pillar could improve the waveguide mode sensitivity (S_{wg}) [57]. By applying SWG of W = 350 nm and t = 300 nm, the simulated S_{wg} shows an improved performance over 0.9 RIU/RIU. However, n_{eff} is not a directly measurable value and requires sensing architectures to convert it into a detectable quantity. For a resonator-based sensor, the sensitivity (S_{res}) consists of the architecture sensitivity (S_{arch}) and waveguide mode sensitivity, which are defined as [57]:

$$S_{\rm res} = S_{\rm arch} \times S_{\rm wg}, \qquad (2.33)$$

$$S_{\rm arch} = \frac{\lambda_{\rm res}}{n_{\rm g}},$$
 (2.34)

and

$$S_{\rm wg} = \partial n_{\rm eff} / \partial \gamma,$$
 (2.35)

¹Parts of the section will be publish: E. Luan, K. M. Awan, K. C. Cheung, and L. Chrostowski, "High performance sub-wavelength grating-based resonator sensors with substrate overetch," *Optics Letters* (accepted). © of OSA. Reprinted with permission.



Figure 2.7: Schematic of the SWG-based waveguide geometry with substrate overetch (Λ is the SWG period, W is the waveguide width, t_{Si} and t_{etch} are the thickness of silicon pillars and BOX pedestals, and n_1 and n_2 represent high and low refractive indices in the SWG waveguide, respectively).

respectively, where ∂n_{eff} is the change of the effective index and $\partial \gamma$ is the variation of any physical parameter surrounding the waveguide. Therefore, not only a large S_{wg} but a small n_{g} are necessary to improve the sensitivity for the resonator-based sensor. Chang et al. developed a pedestal SWG-based resonator sensor, where periodic pillars are semi-suspended onto the SiO₂ substrate by isotropic etching processes [179]. The proposed SWG geometry gives a lower n_{g} and higher index profile symmetricy of the waveguide, thus a larger bulk sensitivity, roughly 545 nm/RIU. However, due to the isotropic etching, the robustness of the sensor is unguaranteed, and moreover, the etching depth (~50 nm) is limited by the width of the SWG pillar, which hinders the potential for further improving sensor performance.

In this section, we introduce a new approach to improve the sensitivity of SWG-based sensors. Instead of etching the substrate isotropically, we used the anisotropic etching process with $Ar/C_4F_8/O_2$ plasma as the etchant, which gives vertical SiO₂ etching down to the buried oxide (BOX) layer and results in the substrate-overetch SWG (SOE-SWG) pillar, as shown in Figure 2.7. This configuration not only enlarged the optical mode to the cladding, but also reduce the group index, both of which will improve sensor performance.

2.3.1 Modelling and simulations

For the design of a microring resonator, several parameters are important to obtain a high optical-performance device, such as the coupling gap distance, the coupling length, the radius of the ring, and the effective index of the bus and resonator waveguides. To obtain the critical coupling for the proposed SOE-SWG MRR sensor, the effective index and propagation loss of the waveguide need to be computed to obtain κ from Equation 2.18. MODE and FDTD Solutions from Lumerical Inc. were employed to calculate the index properties. Then obtained simulation results were introduced to a MATLAB-based script, which is employed analytic functions previously presented by our group [180]. According to the Equation 2.28 and 2.29, the transmission at the through and/or drop ports can be extracted.

Simulations for SWG-based configurations were realized based on Bloch boundary conditions using a fully vectorial three-dimensional Finite-Difference-Time-Domain (3D-FDTD) approach, which has been applied for band structure calculations of periodic structures. To maintain subwavelength-guided wave propagating in the proposed waveguide, we fixed the SWG period of 250 nm, which allows all SWG-based geometries to be investigated within the sub-wavelength regime. Figure 2.8 below shows simulation results of SWG-based waveguides with a varied duty cycle ($\eta = 0.4$ -1) in the TE mode. For SWG waveguides with narrower widths (W = 400 and 450 nm), S_{wg} shows an improved value compared to 500-nm or 550 -nm SWG waveguides (Figure 2.8(a)), which agrees well with the result in Ref. [57]. However, for the resonator-based bulk sensitivity, the maximum value does not appear under the same duty cycle condition, depicted in Figure 2.8(b). That is due to the drop of n_g at the low η region, resulting in a growth of S_{wg} . This indicates that for a resonator-based sensor, the waveguide configuration with the highest S_{wg} does not necessarily have the best S_{res} .

Figure 2.9(a) shows cross-sections of the electric field distribution for SOE-SWG, standard-SWG, and thick-SWG geometries, respectively, in both silicon pillar and gap regions in the TE mode. The duty cycle for each geometry is based on the minimum n_{eff} for a 2-µm-thick BOX layer to minimize the substrate leakage [181]. As the contrast with standard-SWGs (W = 500 nm, $t_{\text{Si}} = 220 \text{ nm}$), both SOE-SWG and thick-SWG geometries show an extended mode profile into the



Figure 2.8: (a) Simulated effective index (left) and waveguide mode sensitivity (right) versus the duty cycle of SWG waveguides with different widths. Three different grayscale regions represent the refractive index of the BOX layer (dark grey), and minimum n_{eff} of SWG waveguides for 3-µm (grey) and 2-µm (light grey) thick BOX layers to minimize the substrate leakage. (b) Simulated group index (left) and bulk sensitivity (right) versus the duty cycle of the SWG waveguides with different widths.

cladding: one is due to the lack of BOX substrate (SOE-SWG); the other one is due to the pillar thickness (thick-SWG). Similar to Figure 2.8, we compared the waveguide properties among these SWG geometries and plotted in Figure 2.9(b) and 2.9(c). $T_{\rm Si}/T_{\rm etch}$ represents the thickness of the silicon pillar and the substrate overetch in each geometry. It is observed that for the thick-SWG geometry (pink region), both $n_{\rm eff}$ and $n_{\rm g}$ increase significantly as the pillar thickness increase. Although S_{wg} also grows (except SWG waveguide with W = 400 nm and $t_{Si} = 500$ nm, which may not support a propagation mode), S_{bulk} presents a downward trend as a function of the pillar thickness. For the SOE-SWG geometry (grey region), the overetch substrate does not influence $n_{\rm eff}$ and $n_{\rm g}$ much (blue curves), but improves the waveguide mode sensitivity, even over than 1 (orange curves above the dashed line in Figure 2.9(b)). which matches the findings in Ref. [182]. It can be explained by the high susceptibility of the SOE-SWG geometry: a slight cladding index variation changes the mode profile resulting in a larger mode effective index change. Therefore, an enhanced S_{bulk} is obtained for the resonator-based SOE-SWG sensor compared to other geometries, as presented in Figure 2.9(c).



Figure 2.9: (a) Cross-sections of the electric field distribution of three different SWG silicon pillars and gaps in the TE mode. (b) Simulated effective index (left) and waveguide mode sensitivity (right), and (c) simulated group index (left) and bulk sensitivity (right) versus different thicknesses of silicon and overetch layers. The grey region represents the SOE-SWG geometry, and the pink region represents the thick-SWG geometry.

2.3.2 Experiments and results

Our SWG-based sensing architectures were fabricated using the EBL system with plasma etching on an SOI platform ($t_{Si} = 220 \text{ nm}$, $t_{BOX} = 2 \mu \text{m}$) by ANT, with fabrication details provided in Ref. [177]. The fabricated SWG MRRs with different waveguide width (W = 400 to 550 nm) and duty cycle ($\eta = 0.4$ to 1) are presented in Figure 2.10. Calibrated transmission spectra for these fabricated MRR with the water cladding on top illustrates that for low duty cycle, due to the small weight ratio of silicon, SWG MRRs show relative low *Q*. Fow *W* of 400 and 450 nm with η of 0.4 (Figure 2.10(a) and 2.10(b)), no resonant peak is observed, and the peak transmission power is lower than -20 dB, which indicates a very high waveguide loss.

In addition, we also characterized sensor performance of fabricated SWG sensors with different W and η . The bulk sensitivity was demonstrated by introducing a series of isopropyl alcohol (IPA, purity \geq 98%, VWR) solutions with different



Figure 2.10: SEM images of the SWG waveguide-based MRRs at the coupling region for the waveguide width of (a) 400 nm, (b) 450 nm, (c) 500 nm, and (d) 550 nm. Each waveguide width has the duty cycle from 0.4 to 1, respectively.

concentrations to sensors through a poly(methyl methacrylate) and polydimethylsiloxane (PMMA/PDMS)-based (SYLGARD 184, Dow Corning) microfluidic gasket above. The gasket was designed through AutoCAD with multilayers, as presented in Figure 2.11(a). The top layer is made of 4-mm-thick PMMA which is diced and drilled by the laser (VLS4.60, Universal Laser Systems). The rest layers consist of 3 layers 0.5-mm-thick PDMS to improve the sealing. Microfluidic patterns were also realized by the laser cutting. The permanent bonding between PMMA and PDMS was according to the Ref. [183], and the rest PDMS layers were bound through the oxygen plasma treatment. The manufactured multilayer PM-MA/PDMS gasket was fixed by using two screws onto the base (Figure 2.11(b)). The temperature was controlled at 25°C during the experiment by a temperature controller (LDC501, Stanford Research Systems).

By sequencing the IPA dilutions from 2% to 8% (v/v) to SWG MRR sensors, the bulk sensitivity of each configuration was characterized. As depicted in Fig-



Figure 2.11: (a) CAD view of the multilayer fluidic gasket for the biotesting.PMMA: top layer, PDMS: 3 bottom layers and the sensing chip: blue (2-channel version). (b) Custom silicon photonic biosensing setup, with the designed PMMA/PDMS 2-channel gasket. Red arrows represent the direction of the solution flow.

ure 2.12, the wavelength shift versus different IPA concentrations is presented for different W and η . Among them, SWG MRRs with η of 100% were measured with no microfluidic system, which present discrete wavelength shift steps (black lines). Since these devices have a minimized distance between sensing rings and grating couplers in order to save the area, the wavelength shift was detected by manually dropping different concentrations of IPA on sensors, and between each concentration, the chip was blown dry by N₂. It is clear to observe that the wavelength shift increases as the duty cycle reduce, which is due to the lower waveguide confinement giving more intensity into the surrounding solutions. At last, the DI water (0% IPA) was introduced again to the sensor, presenting all wavelength shift curves back to the baseline (Figure 2.12).



Figure 2.12: Wavelength shifts as a function of different concentrations of IPA introduced to SWG MRR sensors with the width of (a) 400 nm, (b) 450 nm, (c) 500 nm, and (d) 550 nm. Each waveguide width has the duty cycle from 0.4 to 1, respectively.

Figure 2.13(a) shows the wavelength shift of SWG geometries, with varied *W* and η , as a function of the cladding index change due to different IPA concentrations. The slope of each curve represents the bulk sensitivity, which depicted in Figure 2.13(b). Each bulk sensitivity is also compared with the simulation results. The group index of each geometry was calculated based on $n_g = \lambda^2/(2\pi R \times FSR)$, where *R* is the radius of the ring resonator (*R* = 30 µm). FSRs were measured from transmission spectra around 1550 nm. The measured and simulated group indices are presented in Figure 2.13(c). Based on Equation 2.13, 2.20 and 2.21 ($\partial \gamma$ is the bulk cladding index change), the waveguide mode sensitivity can be also achieved and presented in Figure 2.13(d). Since no obvious resonant peaks for SWG sensors



Figure 2.13: (a) Measured wavelength shift versus the refractive index change in cladding for each sensing architecture. Measured (red curves) and simulated (black curves) (b) bulk sensitivities, (c) group indices, and (d) waveguide mode sensitivities versus the duty cycle of SWG-based sensors with different widths.

with W = 400 and 450 nm and $\eta = 0.4$ were observed due to the low waveguide confinement, thus not presented in Figure 2.13. Compared to the simulation results (black curves), measurements (red curves) show a slightly higher S_{bulk} at the low η region ($\eta = 0.4$ –0.6). That can be explained due to the fabrication imperfection, such as overexposure, which causes stronger influence under low η conditions.

After characterization, the chip was cleaned by the newly prepared piranha solution (3:1 H₂SO₄:H₂O₂) at 100°C for 10 min. Then, a positive-tone photoresist (AZ 5214E-IR, Capitol Scientific) was spin-coated on the chip, and patterned through photolithography, where all SWG MRRs were exposed, but grating couplers and waveguides were protected. The SiO₂ substrate overetch process was realized by using a deep reactive ion etching system (DRIE, SPTS Rapier) with Ar/C₄F₈/O₂ gas at 300:65:35 sccm. The pressure was set to 150 mT, with the in-

ductively coupled plasma (ICP) power of 1000 W and the platen power of 400 W, respectively. The platen temperature was 10°C. To obtain a 300-nm-deep overetch, the chip was etched for 40 sec. At last, the chip was immersed in acetone to remove the photoresist and rinsed with IPA and DI water. After drying, SOE-SWG devices were characterized by using a scanning electron microscope (SEM, Helios NanoLab 650) system as depicted in Figure 2.14(a). It can be observed that approximately 285 nm SiO₂ and 30 nm Si have been etched, respectively, during the SOE process, indicating a SiO₂ to Si etch selectivity of 10:1 (Figure 2.14(b)). However, not only the top surface but also sidewalls around each Si pillar were influenced by the etchant, which turns the SWG block to a trapezoidal object. In addition, the sidewall roughness became worse as compared to the original waveguide in Figure 2.3, which may induce more losses in the resonator. Cross-sectional images of the SOE-SWG waveguide were achieved by using a focused ion beam (FIB) milling process, in the propagation and transverse directions. A $2-\mu$ m-thick platinum (Pt) layer was deposited as the protection layer before the FIB. As shown in Figure 2.14(c), the Si pillar shows a corn-like shape in the propagation direction, with a sidewall angle of 80° , close to the waveguide with no SOE process, as shown in Figure 2.3(b). In the transverse direction (Figure 2.14(d)), the SWG pillar shows a width of 454 nm. Roughly 50 nm Si was etched away.

Transmission spectra of the proposed SWG sensors were re-measured under the water cladding after the SOE process. Based on measurement results, the quality factor and group index were extracted and plotted in Figure 2.15(a) and 2.15(b). In contrast with standard SWG MRRs (blue curves), 300-nm SOE-SWG MRRs (orange curves) show reduced Q and n_g values, which results from the undesired silicon pillar etching during the SOE process. In addition, due to the relatively low etch selectivity, transmission plots for SOE-SWG MRRs with $\eta = 40\%$ and W =400 to 550 nm, and with $\eta = 60\%$ and W = 400 nm present no resonant peaks.

To compare bulk sensitivities, the same IPA dilutions were employed again for SOE-SWG MRR sensor characterizations. Resonant wavelength shifts as a function of the IPA concentration are presented in Figure 2.15(c) for SWG architectures before and after (black) the SOE process, among which a maximum bulk sensitivity is achieved for the SOE-SWG MRR with W = 500 nm and $\eta = 60\%$ ($S_{bulk} = 575$ nm/RIU), roughly 1.2-time of standard-SWG MRRs [83]. According to the



Figure 2.14: SEM images of the SOE-SWG device, with the (a) top-view of the MRR, and the (b) 50-degree-tilted side-view at the coupling region. SEM images with the FIB milled SWG waveguide (W = 500nm, $\eta = 60\%$) cross-sections in the (c) propagation direction, and the (d) transverse direction.

measured Q and n_g in Figure 2.15(a) and 2.15(b), the waveguide mode sensitivity, as well as the intrinsic detection limit (*iDL*, see Euquation 1.10), can be obtained at 1550 nm wavelengths for SWG architectures before and after the SOE process.

As listed in Table 2.1, after the SOE process, sensor performances in terms of the S_{bulk} and S_{wg} are improved for SWG MRRs. Especially for the one with W= 400 and 450 nm, S_{wg} exceeds 1 and is close to the aforementioned simulation result in Figure 2.9(b). However, due to the sharply reduced Q, iDL gets worse to the order of 10^{-3} RIU after the SOE process. At last, a standard bio-sandwich assay was introduced to the SOE-SWG MRR sensor (W = 500 nm and η = 60%) for biosensing demonstration. The detailed bio-sandwich assay is presented as follows. The chip was first introduced with 1 mg/mL protein A for 30 min for the



Figure 2.15: Measured (a) quality factors and (b) group indices versus the duty cycle of SWG waveguides before (blue) and after (orange) the SOE process. (c) Resonant wavelength shift as a function of IPA concentrations. Orange steps (left) and yellow steps (right) represent SOE-SWG rings with $\eta = 60\%$ and 80%, respectively. Black curves are wavelength shifts before the SOE process. (d) Surface bio-sandwich sensing results on the SOE-SWG sensor with W = 500 nm and $\eta = 60\%$.

surface passive adsorption. Then, 20 µg/mL anti-streptavidin (anti-SA), 30 µg/mL bovine serum albumin (BSA), and 30 µg/mL streptavidin (SA) were sequenced into the sensor surface. At last, 50 µg/mL biotinylated-BSA (bBSA) solutions was introduced. Biomolecular interactions with specific and non-specific targets were performed on the proposed sensor surface (Figure 2.15(d)). Each solution injection was followed by a 10-min PSB buffer rinse (grey areas) to remove unbound molecules. Details of the bio-sandwich assay can be found in Ref. [83]. The total wavelength shift for surface interactions is ~900 nm, lower than the former result in Ref. [83] because of the shorter reaction time.

W (nm)	η	S _{bulk} (nm/RIU)		S _{wg} (RIU/RIU)		iLD (RIU)	
		Before	After	Before	After	Before	After
400	60%	514	N/A	0.732	N/A	2.15×10^{-3}	N/A
	80%	348	569	0.786	1.055	$9.46 imes 10^{-4}$	2.01×10^{-3}
450	60%	500	N/A	0.847	N/A	$5.84 imes10^{-4}$	N/A
	80%	345	480	0.835	1.011	$6.59 imes10^{-4}$	1.69×10^{-3}
500	40%	533	N/A	0.719	N/A	$4.15 imes 10^{-3}$	N/A
	60%	479	575	0.905	0.966	$5.48 imes 10^{-4}$	1.75×10^{-3}
	80%	292	411	0.730	0.883	$7.71 imes 10^{-4}$	$1.88 imes 10^{-3}$
550	40%	527	N/A	0.741	N/A	2.45×10^{-3}	N/A
	60%	447	530	0.889	0.990	$5.78 imes10^{-4}$	1.83×10^{-3}
	80%	253	349	0.638	0.917	$5.77 imes 10^{-4}$	2.02×10^{-3}

Table 2.1: Sensor performance of SWG architectures before and after theSOE process at 1550 nm in the TE mode.

2.3.3 Configuration optimization

To improve the quality factor of the proposed SOE-SWG devices, we compensated the etching effect by pre-enlarging the silicon pillar in the as-drawn layout. As SEM images shown in Figure 2.14 and Figure 2.16(a), the SWG pillar present a trapezoidal-like shape after the 300-nm SOE process, which may reduce the effective index of the SWG waveguide. By replacing the rectangle of silicon pillars, as well as BOX pedestals, with pyramids (based on parameters from Figure 2.16(b)), the effective index of the fabricated SOE-SWG waveguide was simulated by using FDTD band structure calculations again. Simulation results indicate that the proposed SOE-SWG architecture based on SEM images has a relatively low effective index of 1.48 with water cladding, which will lead to huge substrate leakage, especially when the chip has 2-µm-thick BOX layer above the silicon substrate. Therefore, to reduce substrate leakage losses, we simulated trapezoidal SOE-SWG pillars with different widths (w) and lengths (l), and kept the height constant at 190 nm, for etching influence evaluation. Simulation results are depicted in Figure 2.17 below. By changing w from 170 to 210 nm, and l from 460 to 500 nm, the effective index of the fabricated trapezodial-like SOE-SWG-based waveguide increases from 1.48 to 1.68.



Figure 2.16: (a) Top view of the SEM image of SOE-SWG configurations with W = 500 nm and $\eta = 60\%$. (b) Schematic of the SOE-SWG configuration with oxide overetch depth of 300 nm. Due to the influence of the etchant, silicon pillar shows a trapezoidal-like shape with the bottom length of 460 nm and the bottom width of 170 nm. Parameters of the trapezoidal pillar are based on previous SEM images.



Figure 2.17: Simulated effective index as a function of the length of SOE-SWG pillars, from 460 to 500 nm, with the width of 170, 190, and 210 nm, respectively, by using FDTD band structure calculations.

Recently, a quality factor highly-improved SWG-based MRR configuration has been reported by introducing asymmetric silicon pillars, which can reduce bend radiation losses and outer sidewall scattering losses by creating an asymmetric effective refractive index profile in the MRR (as shown in Figure 2.18(a)), yielding the Q as high as 11500 with a radius of 5 µm, 4.6 times of that (~2800) offered by a conventional SWG through EBL processes [184, 185]. By utilizing an asymmetric SWG core, an enhanced sensing capability was analyzed and characterized, obtaining a high Q of 9100, bulk sensitivity of 440.5 nm/RIU and surface sensitivity of 1 nm/nm [178], as mention in Chapter 1. Therefore, to improve the Q of the proposed SOE-SWG MRRs, we employed the concept of asymmetric silicon blocks to our configurations. By applying the conformal transformation method, which transforms the effective index profile conformally, a bend SWG-waveguide can be taken equivalently as a straight SWG-waveguide with a transformed Cartesian coordinate system as depicted in Figure 2.18(b). The first order approximation of transformed effective refractive index ($n_{eff}(\rho)$) can be expressed as [186]:

$$n_{\rm eff}(\rho) = \begin{cases} n_{\rm clad} & \rho < r_1, \rho > r_2 \\ n_{\rm core} & r_1 \le \rho \le r_2, \end{cases}$$
(2.36)

and the conformally transformed index value $(n_{con}(u))$:

$$n_{\rm con}(u) = n_{\rm eff}(r_2 e^{u/r_2}) e^{u/r_2}, \qquad (2.37)$$

and

$$u = -r_2 ln \frac{r_2}{\rho},\tag{2.38}$$

where r_1 and r_2 represent inner and outer edges of the MRR. To eliminate the distortion of the effective index profile (shown in Figure 2.18(b)) due to the bend, the pre-distortion compensation was applied by tailoring the SWG pillar top (L_T) and bottom (L_B) lengths. The ideal effective refractive index ($n_{eff}^{ideal}(\rho)$) profile is defined as below [185]:

$$n_{\rm eff}^{\rm ideal}(\rho) = \begin{cases} n_{\rm clad} r_2 / \rho & \rho < r_1, \rho > r_2 \\ n_{\rm core} r_2 / \rho & r_1 \le \rho \le r_2. \end{cases}$$
(2.39)



Figure 2.18: (a) Schematic of a standard SWG waveguide and an asymmetric SWG-waveguide in the MRR. (b) Standard bend waveguide refractive index profile, and conformal transformation method applied refractive index profile corresponding a straight waveguide. Most of the optical mode is confined at the outer side of the bend, thus the inner part index shows a decreased value.

Therefore, $n_{\text{eff}}^{\text{ideal}}(r_1)$ and $n_{\text{eff}}^{\text{ideal}}(r_2)$ are equal to 1.68 and 1.65, respectively, based on the radius of the proposed SOE-SWG MRR sensor of 30 µm. According to FDTD simulations, the asymmetric SWG pillar for the 30 µm radius is achieved with following parameters: $\Lambda = 250$ nm, $\eta = 60\%$, $L_{\text{T}} = 142$ nm, and $L_{\text{B}} = 158$ nm, respectively.

2.4 Conclusion and discussion

In this chapter, we first introduced the figures of merit for a silicon photonic microring resonator, and methods to design an MRR-based biosensor through Lumerical tools and MATLAB. Then, we demonstrated a novel substrate-overetch subwavelength grating-based silicon architecture for sensor performance enhancement. By vertically overetching the SiO₂ between each silicon pillar, the SWG-based waveguide achieves a reduced group index and an enlarged waveguide mode sensitivity simultaneously. Experiment results indicate that the SOE-SWG configuration has improved sensor performance, with a bulk sensitivity as high as 575 nm/RIU. However, due to the low etch selectivity, the etchant also etches the silicon layer (\sim 30 nm) during the 285-nm SOE process, which decreases the thickness and width but increases the sidewall roughness of silicon boxes. This results in higher substrate leakage losses and sidewall scattering losses, both of which reduce the *Q*, thus increasing the iDL. Two approaches can be employed to im-

prove the quality factor. The first approach is compensating the etching effect by pre-enlarging the silicon pillar in the as-drawn layout, and the second approach is applying trapezoidal-shape SWG blocks to reduce the bend radiation losses and outer sidewall scattering.

Chapter 3

Multi-box Sub-wavelength Grating Waveguide Sensors¹

3.1 Introduction

Although we have theoretically and experimentally demonstrated a performanceenhanced SOE-SWG architecture in Chapter 2, the post-fabrication etching processes complex the manufacturing of the sensor. Furthermore, the low etch selectivity between SiO₂ and Si reduces the Q and DL as trade-offs of high sensitivities of SOE-SWG sensors. However, thanks to the state-of-the-art EBL system continues to reduce the minimum Si feature size, performance of Si-based optical sensors can be further improved with no post-fabrication process.

In this chapter, we introduce a novel SWG-based configuration that consists of five rows of periodic silicon squares with 60-nm-gap distances between each square

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(multi-box structure) to construct resonator-based sensors to achieve higher bulk and surface sensitivities. Section 3.2 introduces the design aspects of the multibox SWG waveguide, as well as the simulations for the effective index, losses, and sensor performance. In Section 3.3, experimental measurements are presented for the multi-box SWG geometry with the sensing architecture of microring resonators and phase-shifted Bragg gratings, respectively. Sensor performances are characterized by using a series of refractive index solutions and a standard bio-sandwich assay. In addition, a small molecules interaction model is applied for the proposed sensing architecture for the demonstration of silicon photonic sensor's capability for monitoring interactions with a molecular weight (MW) lower than 1000 Da.

3.2 Theory and simulations

The proposed $180 \times 180 \text{ nm}^2$ periodic silicon square-based multi-box structure with a thickness of 220 nm is presented in Figure 3.1. By periodically repeating Si pillar in both propagation and transverse directions, the multi-box SWG waveguide shows an extended optical mode compared to the conventional Si-core waveguide, with water as the cladding material. The period of the multi-box (Λ) is 240 nm, which includes a Si box (180 nm) and a gap (60 nm). As can be seen in Figure 3.1(e) and 3.1(f), the electric field intensity is mainly condensed between the Si segments, which highly increases the mode overlap with the sensing medium. Moreover, the multi-box structure also enables more surface contact area around each Si segment for the analyte to attach. Therefore, both bulk and surface sensitivities of the device using the multi-box structure will be highly improved compared to other waveguides. Two types of silicon photonic resonator sensors (microrings and Bragg gratings) are proposed and designed in this section by using multi-box SWG waveguides. However, a higher sensitivity usually means a higher loss which influences the limit of detection of a sensor [45]. In the case of the multi-box SWG, losses in waveguides originate from not only the expanded optical mode but also the internal sidewalls of these boxes. In this section, 2D and 3D modelling approaches were applied to study and estimate the loss and sensitivity of the proposed multi-box SWG waveguide.

Theoretically, SWG waveguides are lossless with the light propagating in a



Figure 3.1: (a) Schematic geometry of a 5-row multi-box waveguide in the SOI platform. **(b)** Top view of a 2D representation of the multi-box waveguide. The red $180 \times 180 \text{ nm}^2$ squares represent the Si segments. **(c)** Cross-section view of the multi-box waveguide in the yz-plane. A is the period, and η is the duty cycle. The thickness of the buried oxide (BOX) layer is 2 μ m. **(d)** Cross-section view of the Xy-plane. The height of each segment is 220 nm. **(e)** Cross-section views of the E field distribution in Si segments and gaps in the TE mode with water cladding in the xy-plane.

Bloch-Floquet mode [66]. However, due to the low mode effective index and insufficiently thick buried oxide (BOX) layer, SWG waveguides can exhibit increased substrate leakage loss compared to conventional waveguides [181]. It has been demonstrated that the leakage loss is negligible for BOX thicknesses of 2 μ m when the mode effective index is higher than 1.65, which is independent of the geometric parameters of the SWG waveguide [181]. For the multi-box waveguide sensors, there are extra loss mechanisms: 1) Scattering losses caused by sidewall roughness from the additional interfaces; in the TE mode, most of the field intensity is congregated in the gaps along the x and z directions in Figure 3.1(e), which makes it sensitive to the imperfect sidewall of each Si segment; 2) Material absorption, especially caused by the water cladding at a wavelength of 1550 nm, is the primary optical loss for silicon photonic biosensors (the optical absorption of water is 42.5 dB/cm), since most analytes of interest are found in aqueous solutions.

To evaluate the mode effective index and loss of multi-box SWG waveguides, we applied 2D and 3D simulations as a comparison by using MODE and FDTD

Solutions from Lumerical Inc., respectively. For the 2D simulations, the 3D periodic structure is replaced with an equivalent 2D uniform waveguide by applying the effective medium theory (EMT) along the propagation direction [187]:

$$n_{\rm eq}^2 = \eta n_{\rm Si}^2 + (1 - \eta) n_{\rm clad}^2$$
 (3.1)

where n_{eq} , n_{Si} , and n_{clad} are the RI of the effective medium, the Si and the cladding materials, respectively. For the 3D simulations, a single SWG periodic unit cell is simulated with Bloch boundary conditions as discussed in Chapter 2. It has been successfully applied to simulate the coupling coefficients and effective indices in Bragg gratings and SWG waveguides [83, 188]. In addition to calculating the band structure, this method can also calculate other mode properties, such as group velocity and intrinsic losses [189].

3.2.1 Effective index

Due to the low effective index of multi-box SWG waveguides, the optical confinement is weak, which gives rise to a substrate leakage penalty. To minimize the substrate leakage losses, we studied the effective index of multi-box SWG waveguides with a different number of rows, from 2 to 6, at a wavelength of 1550 nm in the TE mode (all the following simulations are under the same conditions if no specific statement). The structural parameters of each multi-box row are as follows: grating pitch (Λ) = 240 nm, duty cycle (η) = 75%, Si segment width (w) = 180 nm, and waveguide thickness (t = 220 nm). We investigated the waveguide under H₂O and silicon dioxide claddings, respectively. Figure 3.2 presents the effective index of multi-box SWG waveguides as a function of the number of rows by both 2D and 3D simulations. It is observed that 2D simulations (dashed curves) show a lower effective index compared to 3D ones (solid curves) in general. For 5 rows, the effective index has risen to around 1.65, which is the threshold value to ignore the leakage losses with 2-µm-thick BOX [181]. In addition, there is a cut-off index region shown in Figure 3.2, where the effective index value is lower than the RI of the substrate ($n_{SiO_2} = 1.44$) resulting in no guided mode.



Figure 3.2: Mode effective indices of multi-box SWG waveguides with a varied number of rows from 2 to 6 by 2D and 3D simulations. All the simulations are carried out at 1550 nm in the TE mode.

3.2.2 Loss

For the loss analysis, despite the extrinsic loss introduced by fabrication tolerances, we are usually concerned more about the material absorption and bend loss during the design phase. To make sure that there is a sufficient decay (>99%) of the intensity) of the optical field, an 8-µm-wide span simulation region was used from the center of the waveguide with Perfectly Matched Layer (PML), a boundary condition in MODE and FDTD Solutions. Moreover, a mesh override region was selected to force a very small spatial mesh near interfaces for high simulation accuracy. Figure 3.3(a) shows the propagation loss of multi-box SWG waveguides with a different number of rows by 2D (dashed curves) and 3D (solid curves) simulations. For the oxide cladding, the propagation loss decreases with an increase in the number of rows, which is caused by the increased waveguide confinement. Since the optical absorption of silicon dioxide is almost zero, the loss should mainly originate from the substrate leakage and the absorption from the PML. For the water cladding simulations, except the strong loss of 2-row waveguides caused by a low effective index close to the cut-off region, the propagation loss grows gradually. This can be explained by considering the predominant water absorption compared to the loss contribution due to the substrate leakage: the modal area expands corresponding to the growth of the number of rows (3 to 6), which also increases its

overlap area with water. There is a visible difference between 2D and 3D simulations in Figure 3.3(a), which may be caused by the EMT approximation in 2D simulations: the approximation ignores the periodic gap-condensed electric field in the propagation direction, which is fully exposed to the cladding material, resulting in an underestimated loss from the material absorption.

The waveguide bend loss is another important issue that needs to be studied, since it influences the quality factor in a resonator-based device, thus influences sensor performance. It has been reported that for a conventional strip waveguide, the bend loss is negligible at a radius of approximately 10 μ m [190]. In other words, the bend loss can be reduced to the point where it equals the waveguide propagation loss by increasing the bend radius [53]. We modelled and simulated the bend loss of 5-row multi-box SWG waveguides with a varied radius from 5 to 50 μ m by 2D and 3D approaches (the radius is defined from the centre of the waveguide). The bend loss determined in 2D MODE Solutions is generated from the complex eigenvalue. For the 3D simulations, the aforementioned conformal transformation method was applied, by transforming the effective index profile conformally (details can be found in Chapter 2). The effective refractive index profile of 5-row multi-box SWG waveguides with different radii is presented in Figure 3.3(b) after the transformation. A distorted refractive index profile is observed (red curve) which can be treated as a bend with the radius of 5 µm. Comparing to the radius of infinity (black curve), the mode profile of the 5-µm-radius waveguide is dislocated to the outer edge due to the asymmetric index distribution (insets in Figure 3.3(b)). The bend loss analysis can be further performed using 3D-FDTD simulations with Bloch boundary conditions. Figure 3.3(c) shows the simulation results of the 5-row multi-box waveguide with different radii by 2D and 3D simulations, both of which present a strong radius dependence. To further analyze the bend loss, we divided it into the cladding material absorption and bend radiation separately by replacing the cladding with index-customized lossless materials in 2D simulations. Whereas, for 3D bend loss simulations, the absorption loss has already been excluded by applying the transformed effective index profile. In the left y-axis of Figure 3.3(c), the material absorption of water rises slightly from 22.5 dB/cm to 33.4 dB/cm with the radius increasing, whereas for the oxide cladding, the absorption is negligible. In the right y-axis, the simulation results of the bend radiation are presented, comparing to 3D simulations, the radiation loss from 2D ones is still lower due to the EMT approximation.

Additional loss is associated with the coupling of the forward-propagating guided mode with the radiation modes (radiation loss) and the backward-propagating mode (backscattering loss), and is due to surface roughness at the waveguide core boundaries [191]. For the multi-box waveguide, due to its several internal sidewalls, scattering loss caused by the sidewall roughness is not negligible. Studies on slot-waveguide loss [192, 193] show that the scattering loss is around 8 to 10 dB/cm, more than a typical strip waveguide. Hence we assumed our structure should be more sensitive to sidewall roughness. To verify this assumption, in the model we introduced a rough surface element to each Si segment by adding vertical streaks on the sidewall in the propagation direction, and calculated the additional loss compared to smooth-sidewall results. Two parameters that affect the properties of the scattering process are the correlation length (L_c) of disorder, which is the distance between one to another correlated defect [194], and root-mean-square (RMS) roughness (σ). According to the SOI production technology [195, 196], L_c is 50 nm and σ is less than 2 nm. Figure 3.3(d) shows the simulated scattering loss of 5-row multi-box SWG waveguides according to the varied σ from 0 to 3 nm with a fixed L_c of 50 nm. When the RMS value reaches 2 nm, the sidewalls cause a scattering loss of 13.4 dB/cm.

3.2.3 Sensitivity analysis

The bulk sensitivity for resonator-based architectures as a function of the number of rows obtained from the 2D and 3D simulations is plotted in Figure 3.4(a). As can be seen, the bulk sensitivity of multi-box SWG waveguides grows slightly from 3 to 6 rows. This matches well with the trend of modal areas in the inset of Figure 3.4(a): a bigger modal area indicates more overlap with sensing medium, but higher losses (presented in Figure 3.3(a)). For the surface sensitivity, the evanescent field intensity decreases as the growth of the surface adlayer thickness, which creates an effective sensing region ranging from a few to tens of nanometers; exceeding that region will result in a considerable drop of the surface sensitivity [175, 178, 197]. Figure 3.4(b) shows the trend of the surface sensitivity of multi-box SWG waveg-



Figure 3.3: (a) Multi-box waveguide propagation loss as a function of the number of rows with water and oxide claddings respectively by 2D and 3D simulations. (b) Conformal refractive index profile with different bend radii. Inset: The corresponding mode profiles. (c) 5-row multi-box SWG waveguides material absorption and bend radiation loss versus radius by 2D and 3D simulations. (d) Scattering loss of 5-row multi-box SWG waveguides by 3D simulations versus different RMS values from 0 to 3 nm ($L_c = 50$ nm).

uides according to the increase of well-distributed protein adlayers. For a surface with optimum occupation by proteins, the refractive index is $n_{add} = 1.48$. Both the 2D and 3D simulations show a linear sensitivity below 30 nm of the adlayer, after which the sensing performance decays dramatically. Since most of the optical field is highly condensed in the 60 nm gaps, the sensitivity of the top surface of the sensor is reduced [178]. Thus, when gaps are filled, increasing the thickness of adlayer will not change the RI much. For consideration of the simulation time and memory requirements, we use different mesh sizes for 2D and 3D surface sensitivity simulations separately. The mesh size can be scaled down to 1 nm for 2D simulations for accuracy, whereas, to avoid the heavy computation of geometry sweeps,



Figure 3.4: (a) Bulk sensitivity of the multi-box waveguide as a function of the number of rows by 2D and 3D simulations. Inset: Modal area of each waveguide with water cladding by using MODE Solutions. (b) Surface sensitivity of the multi-box waveguide with varied protein adlayer thickness by 2D and 3D simulations. The mesh sizes are 1 nm and 5 nm for 2D and 3D simulations, respectively.

the mesh size for 3D simulations is set to 5 nm. Moreover, based on the lower effective index by 2D simulations in Figure 3.2, the optical mode should be wider compared to 3D ones, which explains the higher bulk and surface sensitivities by 2D simulations presented in Figure 3.4(a) and 3.4(b).

3.2.4 Temperature stability

The waveguide thermo-optic coefficient (TOC) is dependent on the materials. Thus, different ratios of cladding to Si in the waveguide will influence the temperature sensitivity of the device. With the material TOCs given for silicon, silicon oxide and water ($TOC_{SiO_2} = 2.8 \times 10^{-5}/K$ [53], $TOC_{Si} = 1.87 \times 10^{-4}/K$ [53], and $TOC_{H_2O} = -9.9 \times 10^{-5}/K$ [198]), the refractive index of the water cladding decreases when the environment temperature rises, which is opposite to silicon and silicon oxide. As a result, the thermal-based change of the effective index with water cladding is smaller as compared with the oxide cladding, and leads a reduced temperature sensitivity. Conventional strip waveguides are the most temperature-sensitive structures due to the high Si percentage and strong confinement in the cross-section. With the decrease of the Si proportion in the core waveguide from strip to SWG, more light propagates in the cladding layer, which effectively min-

imizes the shift of resonance wavelengths. Therefore, multi-box SWG waveguide geometry with water cladding should present a lower temperature sensitivity as compared to the conventional strip waveguide.

3.2.5 Sensing architecture

Two multi-box SWG resonator-based architectures are applied for biosensing applications in this section: microring resonators (MRR), and phase-shifted Bragg grating resonators (PSBG). For the multi-box SWG microring resonator-based sensor, as seen in Figure 3.5, both the bus waveguide and the ring are made of multi-box periodic Si blocks in the coupling region. That is based on the consideration to maintain a strong coupling strength, especially for point coupling: the coupling strength is much smaller in asymmetric waveguides as compared with the full coupling in the case of symmetric waveguides [199]. A 20-µm-long taper is employed to convert the standard waveguide to the multi-box SWG waveguide. Details about designing the MRR architecture can be found in Chapter 2. However, MRR sensors have disadvantages such as the large silicon footprint and high bend radiation losses, compared to Bragg gratings based resonators. Therefore, in this part, we will describe more details about the phase-shifted Bragg grating resonator design with multi-box SWG configurations for sensor performance enhancement.

Figure 3.6 below illustrates the schematic of the multi-box Bragg grating device with a phase-shifted cavity at the center. Two periods exist in this configuration: one is the period of Bragg gratings (Λ_1); the other is the period of the multi-box SWG ($\Lambda_2 = \Lambda_1/2$). RI modulation for the Bragg grating is achieved by periodically adding corrugation boxes parallel to the main waveguide, with a corrugation width of ΔW . The length of each corrugation box is the same as the silicon box (L_{Si}) inside the main waveguide, which equals $\eta \times \Lambda_2$ (η is the duty cycle of the multi-box SWG). On each side of the phase-shifted cavity, the Bragg unit is replicated *N* times as the reflector.

The stop band and resonant peak in the transmission spectrum can be optimized by adjusting various grating parameters through design. Simulations of multi-box Bragg gratings were realized by employing Bloch boundary conditions on a single grating periodic cell (as shown in Figure 3.6) using the 3D-FDTD band structure


Figure 3.5: (a) Schematic of a multi-box microring resonator and a multibox coupling waveguide. (b) The magnified image of the black dashed box; Λ is the multi-box period, and the length of each Si segment is determined by the duty cycle η . (c) The magnified waveguide crosssection showing the 5-row of multi-box segments with a thickness of 220 nm exposed in a sensing medium. The model is not in scale.

calculations. By sweeping the length of the silicon box ($L_{Si} = 180-200$ nm) but keeping the gap distance constant ($L_{gap} = 60$ nm), the effect of Bragg period ($\Lambda_1 = 2(L_{Si} + L_{gap}) = 480-520$ nm) on the central wavelength position of the stop band is shown in Figure 3.7(a) with the width of corrugation boxes fixed ($\Delta W = 120$ nm). All band structure calculations assume that the periodic cell is infinitely repeated ($N = \infty$).

We used a coupled-mode-theory-based transfer matrix method (CMT-TMM) to calculate the spectral responses of proposed multi-box PSBGs with various geometrical parameters. The transmission and reflection responses can be calculated using [200]:



Figure 3.6: Two-dimensional schematic geometry of a 5-row multi-box PSBG waveguide with a thickness of 220 nm (Λ_1 and Λ_2 are the period of Bragg gratings and multi-box blocks, L_{Si} and L_{gap} are the length of silicon boxes and gaps in between, *W* and ΔW are the width of the main waveguide and corrugation boxes, and *N* is the number of periods on each side).

$$\begin{bmatrix} A_{\rm in} \\ B_{\rm in} \end{bmatrix} = GPG \begin{bmatrix} A_{\rm out} \\ B_{\rm out} \end{bmatrix} = \begin{bmatrix} M_{11} & M_{12} \\ M_{21} & M_{22} \end{bmatrix} \begin{bmatrix} A_{\rm out} \\ B_{\rm out} \end{bmatrix}, \qquad (3.2)$$

where A and B are the electric fields of the forward and backward propagating modes at the input and output of the device, respectively, G represents the transfer matrices of the grating on each side, and P represents the matrix of the phase-shifted cavity in the middle. The analytic solution for the matrix G can be obtained by treating it as a uniform grating:

$$G = \begin{bmatrix} \cosh(sL_{\rm G}) + j\frac{\Delta\beta'}{2s}\sinh(sL_{\rm G}) & j\frac{\kappa}{s}\sinh(sL_{\rm G}) \\ -j\frac{\kappa}{s}\sinh(sL_{\rm G}) & \cosh(sL_{\rm G}) - j\frac{\Delta\beta'}{2s}\sinh(sL_{\rm G}) \end{bmatrix}, \quad (3.3)$$

where $s = \sqrt{\kappa^2 - (\Delta \beta'/2)^2}$, $\Delta \beta' = 2\beta' - 2\pi/\Lambda$, and $\beta' = \beta - j\alpha/2$. Among them, β is the propagation constant in the grating, Λ is the period of Bragg, L_G is the length of the grating mirror on each side, κ is the coupling coefficient, and α is the propagation loss per length, respectively. For the phase-shifted cavity, the central matrix *P* is given by:

$$P = \begin{bmatrix} e^{j\beta' L_{\rm P}} & 0\\ 0 & e^{-j\beta' L_{\rm P}} \end{bmatrix}, \qquad (3.4)$$

where L_P is the length of the central phase-shifted cavity. Therefore, the transmission spectrum of the through-port is given by $T = 1/|M_{11}|^2$.

We chose the silicon box length of 190, 185, and 180 nm for 3, 4, and 5row multi-box Bragg gratings, respectively, and evaluated the Q of each PSBG configuration as a function of the corrugation width. The propagation constant β of the excited Bloch mode in the grating was calculated through $\beta = 2\pi \times n_{\text{eff}}/\lambda$. The coupling coefficient κ can be extracted by [53]:

$$\kappa = \frac{\pi n_{\rm g} \Delta \lambda_{\rm PSBG}}{\lambda_{\rm c}^2},\tag{3.5}$$

where λ_c is the central wavelength of $\Delta\lambda_{PSBG}$. The propagation losses α of 3, 4, and 5-row multi-box SWG waveguides (with no corrugations) were approximately 35, 40, and 45 dB/cm, respectively, which are based on our simulations from Figure 3.3(a) [92]. By bringing β , κ , and α into the above formulas, a transmission spectrum is acquired for each PSBG, where $L_G = N \times \Lambda_1 = 140\Lambda_1$ and $L_P = \Lambda_1/2$ (defined as the quarter-wave PSBG). Simulated Q are extracted directly from transmission spectra and plotted in Figure 3.7(b). By enlarging ΔW from 80 to 200 nm, a growing Q value is observed, which results from the increased κ and decreased mirror loss (α_{mirror}). However, it is important to note that Q is ultimately limited by the waveguide loss (α_{wg}); in other words, the maximum transmission is not unity due to the loss, which determines the extinction ratio (ER) of the resonant peak [55].

3.3 Experiments and results

3.3.1 Design and fabrication

Devices were fabricated on an SOI wafer with 220 nm thick top silicon layer and 2 μ m BOX layer (SOITec, Grenoble, France) by using the NanoSOI MPW fabrication process by Applied Nanotools Inc. (Edmonton, Canada). The patterning



Figure 3.7: (a) Simulated central wavelengths of the stop band as a function of the length of multi-box blocks $(L_{\rm Si})$ for 3, 4, and 5-row multi-box Bragg gratings ($\Delta W = 120$ nm, and $L_{\rm gap} = 60$ nm). (b) Simulated Q from the transmission spectra versus the variation of the corrugation width of 3, 4, and 5-row multi-box Bragg gratings ($L_{\rm Si-3row} = 190$ nm, $L_{\rm Si-4row} = 185$ nm, and $L_{\rm Si-5row} = 180$ nm).

process begins by cleaning and spin-coating the EBL resist on the SOI wafer; then, a 100 keV electron-beam lithography (EBL) system is used to define the device pattern into the resist; after the chemical develop, an anisotropic inductively coupled plasma reactive-ion etching (ICP-RIE) process is performed on the substrate to transfer the pattern to the underlying silicon layer. The minimum feature size is 60 nm [201].

To experimentally validate the simulation results of the loss, a set of multi-box straight waveguides with a different number of rows and 5-row multi-box bend waveguides with different radii were repeated on two SOI chips. A 2.2 μ m silicon dioxide upper cladding was deposited by plasma enhanced chemical vapor deposition (PECVD) on one of the chips as a comparison. Moreover, for sensing applications, multi-box waveguide-based microring and phase-shifted Bragg grating resonators were also fabricated with no oxide cladding on top.

3.3.2 Optical measurements

All the devices were characterized by an automated optical test setup. A tunable off-chip laser source (Agilent 81682A, with a range from 1460 to 1580 nm) with

power meters (Agilent 81635A) was used for measuring the transmission spectra. An optical fiber array (PLC Connections, Columbus, OH) with four polarizationmaintaining fibers was applied to couple the light on and off the chip. The stage was thermally controlled by a temperature controller (SRS LDC501) to avoid thermal induced resonance shift. The aforementioned gasket with 2 parallel 300-µmwide channels and a syringe pump (Chemiyx Nexus 3000) were integrated over the optical sensor for sensing detection. Figure 3.8(a) shows the SEM images of the multi-box SWG waveguide-based MRR sensors, with the row number of five. The radius of the MRR is 30 μ m, and a 20- μ m-long taper is used to slightly convert the standard waveguide to multi-box SWG waveguides. For the Bragg grating architectures, 3, 4, and 5 rows PSBGs were fabricated in a testing structure consisting one input port and three output ports, as shown in Figure 3.8(c). Similar to the MRR, PSBG also uses the 20-µm-long taper for the waveguide conversion. The measured transmission spectra of proposed resonators are plotted in Figure 3.8(b) and 3.8(d), showing an estimated Q ($\sim \lambda_{res} / \Delta \lambda_{FWHM}$) of 2600 with the extinction ratio (ER) of 29 dB and free spectral range (FSR) of 6 nm for the microring, and a Q as high as 7800 with the ER of 23 dB for the Bragg grating (3-row PSBGs).

3.3.3 Loss measurement

To obtain the multi-box waveguide loss from experimental measurements, a varied length from 0 to 1000 μ m of straight waveguides and a varied cascade from 2 to 20 of bend waveguides were fabricated and tested in the TE mode. Figure 3.9 shows the measured losses at 1550 nm in straight and bend multi-box SWG waveguides with water and oxide claddings separately. The propagation loss in straight waveguides is presented in Figure 3.9(a) as a function of the number of rows. The overall trend of the actual measurement results is similar to the simulations in Figure 3.3(a), but the measurements show roughly 13 dB/cm higher than the simulations, which matches well with the simulated scattering loss due to the sidewalls. From 3 to 5 rows, the propagation loss increases correspondingly not only due to the extended modal area but the total increased sidewall numbers. Figure 3.9(b) presents the experimentally measured loss per 90° bend with different radii from 5 to 50 µm as well as the propagation loss with water (black solid line, $\alpha_{water} = 48.2$



Figure 3.8: (a) SEM images of a 5-row multi-box microring with a radius of 30 μ m. (b) Measured transmission spectrum of the multi-box microring with FSR = 6 nm, Q = 2600 and ER = 29 dB. (c) Schematic of the proposed sensing device with one input and three outputs connecting with 3, 4, and 5-row multi-box PSBGs, as well as the scanning electron microscope (SEM) images, focused at the center of the phase-shifted cavity, where silicon is false-colored. (d) Measured transmission spectra of the proposed device, after calibrated by a similar 4-port device where PSBGs are replaced by standard waveguides.

dB/cm) and oxide (black dashed line, $\alpha_{oxide} = 13.3$ dB/cm) claddings. Moreover, simulation results are also added in Figure 3.9(b) as a comparison, in which 2D simulations including material absorption and bend radiation are discrepant to the measurements due to the scattering loss.

3.3.4 Sensitivity characterization of rings

We characterized sensor performance of multi-box SWG waveguide-based sensors with MRR and PSBG sensing architectures, respectively. In this part, we fo-



Figure 3.9: (a) Measured and simulated straight multi-box waveguide loss as a function of the number of rows from 2 to 5, and (b) bend 5-row multi-box bend loss as a function of the radius from 5 to 50 μ m at a wavelength of 1550 nm in the TE mode with water and oxide claddings ($\Lambda = 240$ nm, $\eta = 75\%$ and t = 220 nm with 2 μ m thick BOX).

cused on the sensitivity characterization of microring resonator-based sensors. To compare the sensitivity of the proposed multi-box SWG structure with other wellstudied architectures, microring resonator-based architectures were selected, which include the 5-row multi-box SWG MRR, the 750-nm-wide MRR in the TM mode and the conventional SWG waveguide-based MRR. Isopropyl alcohol (IPA) is soluble in water and volatile at room temperature, no residues are left on the silicon chip after rinsing and drying by nitrogen. A set of IPA (JT Baker, 4L) with different concentrations ranging from 5% to 20% (v/v) was employed as the refractive index standards for bulk sensitivity measurements. The stage is thermally tuned to 25°C during the optical measurement to minimize the impact of external thermal noise and drift. Each MRR sensor was measured multiple times for each concentration to ensure the stability and accuracy of the signal. Figure 3.10(a) shows the steps of resonant wavelength shifts on three different microring structures. The slopes of the wavelength shift per change of RI provide the bulk sensitivities, which are $S_{\text{TM}} = 228 \text{ nm/RIU}, S_{\text{SWG}} = 387 \text{ nm/RIU} \text{ and } S_{\text{multi-box}} = 580 \text{ nm/RIU}, \text{ respectively}$ (Figure 3.10(b)). According to the measured Q in Figure 3.8(b) and $S_{\text{multi-box}}$, we can achieve an iLD_{bulk} of 1.02×10^{-3} RIU for the multi-box microring resonator.

Similarly, based on the measured Q values of the 750 nm TM mode and SWG waveguides based resonators ($Q_{\rm TM} = 10778$, $Q_{\rm SWG} = 5270$), we also calculated bulk iDL values of 6.3×10^{-4} RIU and 7.6×10^{-4} RIU respectively.



Figure 3.10: (a) Measured peak wavelength shifts of three resonators: the TM mode microring (w = 750 nm, t = 220 nm), SWG microring (w = 500 nm, t = 220 nm, $\Lambda = 250$ nm, $\eta = 70\%$) and 5-row multi-box microring (w = 1200 nm, t = 220 nm, $\Lambda = 240$ nm and $\eta = 75\%$) in the TE mode. (b) Calculated bulk sensitivity results of each resonator at 25° C.

The surface sensing performance of the multi-box microring resonators was evaluated by using a standard sandwich assay to probe biomolecular interactions with specific and non-specific targets. As illustrated in Figure 3.11(a), the sensor was rinsed with PBS buffer (pH = 7.39, n = 1.35) for 20 minutes to achieve a baseline at 37°C before any functionalization. Protein A (0.5 mg/mL, MW ~42 kDa, ThermoFisher, Chicago, IL), a globular protein with a diameter of 3 nm [202] and a refractive index of 1.48 [203], was first introduced to the sensor's surface for irreversible binding as shown in Region A. This process has been proven to facilitate the immobilization and orientation of the capture antibodies [204, 205]. After passive adsorption of Protein A, the capture antibody, anti-Streptavidin (antiSA, 20 µg/mL, MW ~150 kDa, Vector Labs, Burlingame, CA) is introduced and bound to the Protein A adlayer for sensor functionalization (Region B). Then, to prevent unwanted adsorption to the sensor's surface, Bovine Serum Albumin (BSA, 30 µg/mL, MW ~66 kDa) was introduced to block any remaining exposed surface

sites, as shown in Region C. Next, the functionalized and blocked sensor was subjected to Streptavidin (SA, 20 µg/mL, MW ~57 kDa, Vector Labs, Burlingame, CA) presented in Region D, which can be bound by the antibody specifically and irreversibly. In the final step, biotin conjugated with BSA (50 µg/mL, MW ~66 kDa, Vector Labs, Burlingame, CA) was introduced as an amplification step, and since biotin and SA have one of the strongest non-covalent binding interactions [206], another permanent resonant shift could be observed in Region E. Each reagent was followed by a 30 min PBS buffer rinse to remove any unbound molecule on the sensor's surface (blue area labelled 'PBS' in Figure 3.11(b)) when they were injected over the sensor sequentially.

Figure 3.11(b) shows the peak wavelength shifts of the biological sandwich assay on three different microring resonator devices; TM mode, SWG and 5-row multi-box. Comparing with the TM and SWG microring resonators, the multibox waveguide-based sensor shows a great enhancement in each wavelength shift caused by different protein layers added on the surface. It is reported that Protein A can form a physisorbed layer of approximately 1-3 nm thick [20, 205]. However, in reality, the surface coverage of the protein adlayer is not 100%; a constant refractive index of the protein layer ($n_{\text{protein}} = 1.48$) with changing effective thickness is selected as a thin layer simulation. Assuming this variation is linear [203], our simulation indicates that a 1 nm layer needs to cover 42% of the sensor's surface to result in the observed wavelength shift for the multi-box device in Region A. Likewise, a 3 nm layer needs to cover 14% of the surface to lead to a similar wavelength shift. If an antibody can be modelled as a 5 nm diameter cylinder with 10 nm height [83], the shifts in Region B suggest a 1.4 nm uniform adlayer, or approximately 14% surface coverage, which is also well agreed with the previous surface coverage values for antibodies [20, 83, 205]. There is no obvious shift in Region C, which may be due to the surface saturation after introducing the antibody resulting in no potential sites for further binding of BSA. In Region D and E, the wavelength shift of Biotin-BSA is approximately 2X higher than the shift of SA, which indicates that not all four binding sites are occupied with Biotin-BSA, although SA is tetravalent. There is a slight tilt observed in peak-shift curves which represents a slow blueshift happening simultaneously during the assay. This phenomenon has been reported and interpreted as the Si oxidization by water [207].

The total thickness of captured protein multi-layers is estimated to be around 15 to 30 nm [20]; considering the situation of 14% surface coverage, the effective thickness is between 2.1 to 4.2 nm. Based on the overall wavelength shifts presented in Figure 3.11(b) ($\Delta\lambda_{multi-box} = 6000 \text{ pm}$, $\Delta\lambda_{SWG} = 3000 \text{ pm}$ and $\Delta\lambda_{TM} = 1000 \text{ pm}$), surface sensitivities of 1900, 950 and 310 pm/nm are observed for the multi-box, SWG and TM microring resonators, which agree well with the simulation results in Figure 3.4(b). We can further obtain the values of surface iDL of the three measured structures, which are 3.13×10^{-1} nm for multi-box, 3.26×10^{-1} nm for SWG, and 4.64×10^{-1} nm for TM microring resonators.



Figure 3.11: (a) Schematic of standard biological sandwich bioassay: Region A = Protein A (0.5 mg/mL), B = anti-streptavidin (antiSA) (20 µg/mL), C = Bovine Serum Albumin (BSA) (30 µg/mL), D = streptavidin (SA) (20 µg/mL), and E = biotinylated-BSA (50 µg/mL). The model is not in scale. (b) Real-time biosensing experimental results of the resonance shift for the TM (w = 750 nm, t = 220 nm), SWG (w = 500 nm, t = 220 nm, $\Lambda = 250$ nm, $\eta = 70\%$) and multibox (w = 1200 nm, t = 220 nm, $\Lambda = 240$ nm and $\eta = 75\%$) microring resonators during the bio-sandwich assay. The sensor is washed with PBS for 30 min after each reagent presented in blue area.

Moreover, we also characterized the temperature sensitivity of MRR sensors with three different waveguide geometries in the TE mode. The temperature of the stage was controlled by the thermal controller and ranged from 25 to 28°C during the injection of DI water above the devices. The flow rate was regulated at 1 μ L/min to minimize the temperature difference between the waveguides and water cladding. Figure 3.12 presents the resonant wavelength shifts of the three

resonators for different temperatures. Due to the lowest Si ratio in the waveguide core, the multi-box SWG microring resonator shows a blue shift when increasing the temperature ($S_{multi-box} = -30 \text{ pm/K}$), whereas for the TM mode and SWG resonators, red shifts were observed ($S_{TM} = 28 \text{ pm/K}$ and $S_{SWG} = 25 \text{ pm/K}$). Therefore, it is possible to design a thermally independent waveguide geometry with water cladding if the portion of water and silicon in the waveguide is fine-tuned.



Figure 3.12: Resonant wavelength shifts caused by different temperatures of three resonators: the TM mode microring (w = 750 nm, t = 220 nm), SWG microring (w = 500 nm, t = 220 nm, $\Lambda = 250$ nm, $\eta = 70\%$) and 5-row multi-box microring (w = 1200 nm, t = 220 nm, $\Lambda = 240$ nm and $\eta = 75\%$) in the TE mode.

3.3.5 Sensitivity characterization of Bragg gratings

As shown in subsection 3.3.4, both bulk and surface sensitivities of the multi-box SWG waveguide are enhanced compared to TM-mode, or SWG waveguide-based sensors. However, in terms of the detection limit, due to a relatively low Q (~2600) of the multi-box SWG-based MRR sensor, iDL of the bulk sensing is only on the order of 10^{-3} RIU. Therefore, we introduce the phase-shifted Bragg grating resonator as the multi-box SWG-based sensing architecture in this subsection to improve the performance of DL. To have a better comparison, we designed and

fabricated PSBG sensors with a varied row number (3, 4, and 5 rows in the main waveguide).

The bulk sensitivity was assessed by injecting NaCl dilutions ranging from 62.5 to 500 mM through sensing devices, with a flow rate of 20 μ L/min, on the stage thermally tuned to 25°C. Refractive indices of NaCl dilutions (n_{NaCl}) were calculated at 1550 nm wavelengths based on a third-order polynomial fit [208]:

$$n_{\rm NaCl} = -0.008w^3 + 0.074w^2 + 0.162w + 1.3162, \tag{3.6}$$

where *w* indicates the weight fraction of NaCl dilutions within the solubility range (*w* = 0-0.25). As shown in Figure 3.13(a), the sensorgrams of 3, 4, and 5-row multibox PSBGs present stepped wavelength shifts as a function of the concentration of NaCl, leading to a bulk sensitivity of 551.2, 567.5, and 579.2 nm/RIU, respectively (Figure 3.13(b)). Experiments were repeated multiple times to ensure signal accuracy and stability. To further determine the detection limit (sDL) of proposed sensors, the system noise floor, which includes the thermal drift, light source fluctuation and detector noise, was evaluated by constantly introducing deionized (DI) water to sensors for 3 hours, obtaining the three standard deviations (3 σ) of 2.3, 3.0, and 3.4 pm, respectively. In terms of bulk sensing, the system DL (sDL) is on the order of 10⁻⁶ RIU according to the Equation 1.5. As for the intrinsic detection limit, the lowest iDL of proposed multi-box PSBG sensors is around 3.6 × 10⁻⁴ RIU achieved from the 3-row PSBG sensor, comparable to the best resonator-based SOI sensors [209].

To investigate the surface sensitivity, a layer-by-layer electrostatic polymer deposition approach was employed based on Ref. [175] by introducing polyethylenimine (PEI), polystyrene sulfonate (PSS), and polyallylamine hydrochloride (PAH) to the sensor surface. All polymers were dissolved in Tris buffer (0.5 mM, pH = 7.4, Sigma-Aldrich) to 5 mg/mL. As illustrated in Figure 3.14(a), the sensor chip was first cleaned by a piranha solution (3:1 H₂SO₄:H₂O₂) at 100°C for 10 min, which removes organic pollutants and forms hydroxyl groups on the surface (caution, piranha solution reacts violently with organic solvents). Then, the chip was exposed to positively charged PEI (MW 5000 Da, Sigma-Aldrich) for 5 min to ensure the sufficient coverage of the initial adhesion, and followed by adequate Tris



Figure 3.13: (a) Wavelength shift sensorgrams of bulk refractive index (RI) steps with NaCl dilutions from 62.5 to 500 mM. Inset: The system noise floor of proposed multi-box PSBG sensors at 25°C. (b) Bulk sensitivity results of 3, 4, and 5-row multi-box PSBGs.

buffer rinse. Next, negatively charged PSS (MW 70000 Da, Sigma-Aldrich) and positively charged PAH (MW 17500 Da, Sigma-Aldrich) were alternated to the surface for 5 min, respectively. A 5-min Tris buffer flushing was performed after each deposition to avoid polymer precipitation and clogging. All solutions were injected with a flow rate of 20 µL/min at 25°C. The layer-by-layer deposition result of polyelectrolytes is presented in Figure 3.14(b) as a function of time. The inset of Figure 3.14(b) shows a zoom-in view of two successive deposition cycles. To obtain the surface sensitivity of proposed multi-box PSBG sensors, the thickness of the bilayer is required. Glass slides with 5, 10, and 15 bilayers were manually prepared by using the same polyelectrolytes, and measured through a thin-film measurement system (Filmetrics F20-UVX) with an assumption of $n_{polymer} = 1.50$ at 1550 nm wavelengths. Measurements indicate a thickness of 1.987 nm per PSS/-PAH bilayer depicted in the inset of Figure 3.14(c). Surface sensitivities of multibox PSBGs are presented in Figure 3.14(c) with a highest S_{surf} of 1941 pm/nm, in good agreement with previous experiments [92].

Real-time measurements of protein molecule interactions are of great importance for both fundamental research and biomedical applications. Due to the large variety of chemical compounds from natural or pharmaceutical sources, the detection and quantification of small molecules (typically less than 1000 Da in size)



Figure 3.14: (a) Schematic of the layer-by-layer polyelectrolytes deposition process at the surface of proposed sensors. (b) Measured wavelength shifts of 3, 4, and 5-row multi-box PSBGs in terms of the deposition of polymers. Inset: An amplification of the wavelength shift in two deposition cycles: a more substantial shift (2 nm) is observed in PSS than in PAH (1 nm) due to the molecular weight (MW) difference. (c) Surface sensitivities of proposed sensors. Inset: Measured thicknesses of different numbers of PSS/PAH bilayers on the glass slide.

have attracted much attention in a wide range of fields [210]. Surface plasmon resonance (SPR) has been demonstrated as one of the most powerful technologies to determine specificity, affinity and kinetic parameters for label-free protein-protein binding analysis [211]. Due to the increasing demand for modern drug discovery approaches, the development of high-performance SPR technology is further accelerated, prompting SPR as a validated tool for monitoring small molecule interactions [212]. However, SPR-based approaches usually require expensive and bulky instruments as well as trained operators, which restricts their broad adoption for the clinical and home health-care diagnosis, especially in developing countries. Consequently, silicon photonic-based planar sensors are regarded as a simplified and disposable alternative. However, the magnitude of the wavelength shift or intensity change for an evanescent field-based sensor is linearly proportional to the mass change due to the adsorption at the surface, it is challenging for the conventional waveguide-based silicon photonic sensor to detect analytes with small MWs, typically at low concentrations.

To explore the small molecule monitoring capability of the proposed sensor, we applied the well-studied biotin-streptavidin model to the 3-row PSBG sensing device. Biotin is a water-soluble B-vitamin with a very small MW of 244 Da. It has the strongest known non-covalent protein-ligand interaction with the tetrameric protein avidin (also streptavidin and neutravidin), showing a dissociation constant (K_d) of 10^{-15} M (molar) in solution [213]. Instead of the multi-cycle kinetics (MCK) assay which is commonly used in SPR detection, the kinetic titration series (or single-cycle kinetics, SCK) approach was employed in this work [214]. Different from the classic MCK requiring complete analyte removal between measurement cycles, SCK involves sequentially injecting an analyte concentration series without any regeneration steps, which provides binding constants as precise as the MCK method of analyte injections [215].

In comparison with direct adsorption, proteins attached via chemical links at the sensor surface are in their natural globular conformations and provide high activity [216]. Therefore, as depicted in Figure 3.15(a), the sensor surface was functionalized through a chemical linking method: after cleaning with the piranha solution, the chip was first exposed to phosphate buffered saline (PBS, pH = 7.4, Gibco) to obtain a stable baseline; then, a high concentration (150 µg/mL) of biotinylated Bovine Serum Albumin (bBSA, MW ~66 kDa, Vector Labs) was introduced to the surface as a crosslinker (Step A), and incubated for 1 h for a near-maximum coverage; after coating with bBSA, 50 µg/mL streptavidin (SA, MW ~57 kDa, Vector Labs) was followed up for the surface modification via the biotin-SA linkage (Step B); at last, the chip was subjected to 50 µg/mL Bovine Serum Albumin (BSA, MW ~66 kDa, Sigma-Aldrich) to block any remaining exposed surface sites (Step C). Each step was followed by copious PBS buffer rinse to remove any unbound molecule on the surface. After the attachment of bio-receptors, a set of biotin dilutions with concentrations of 10^{-11} , 10^{-9} , 10^{-7} , 10^{-5} , and 10^{-3} M were injected over the sensor sequentially (Step D). The association and dissociation times were set to 5 min and 3 min, respectively, with a flow rate of 30 μ L/min for each concentration. All steps were performed at 25°C. In addition, a control experiment was performed in parallel to quantify the specific binding between biotin and SA molecules, where all the surface modification steps were the same except that SA is replaced by BSA in Step B. The sensorgrams of surface modification and SCK assay are shown in Figure 3.15(b), presenting the time-dependent wavelength shift for both sensing and control experiments.

The wavelength shift of the biotin-SA interaction is depicted in Figure 3.15(c), presenting saw-tooth like patterns after baseline detrending. For the sensing data (blue curve), the resonant peak shows 0.08, 1.51, 7.02, 10.05, and 10.97 pm shifts for each concentration of biotin dilutions. As for the reference data (red curve), there is no noticeable shift larger than the $3\sigma_{3row}$ of the systematic noise after PBS rinsing, indicating a negligible non-specific binding between biotin and BSA. The analyte binding capacity of the sensing surface depends on the density of available binding sites from immobilized receptors. The theoretical maximum binding response R_{max} of the analyte is determined by [217]:

$$R_{\rm max} = \frac{M_{\rm analyte}}{M_{\rm receptor}} R_{\rm ads} \times n \tag{3.7}$$

where M_{analyte} and M_{receptor} are MW values of the analyte and receptor, respectively, R_{ads} is the sensor response of the immobilization of receptors at the surface, and *n* is the molar ratio of the binding sites in the receptor. Due to the surface immobilization of the SA capture protein, it is predicted that two (*n* = 2) of the four potential biotin-binding sites on SA will be available for subsequent interactions with soluble biotin [218]. The adsorption of 50 µg/mL SA (Step B) presented an observed resonant wavelength shift of $R_{\text{ads}} = 1.33$ nm following the buffer rinse (Figure 3.15(b)). Based on Equation 3.7, the device's theoretical R_{max} of the biotin-SA interaction for two binding sites is 11.39 pm, remarkably close to the observed value of 10.97 pm.

The resonant wavelength shift as a function of the concentration of biotin is plotted in Figure 3.15(d), and fitted with the Hill-Langmuir equation $\Delta\lambda/\Delta\lambda_{max} = c^n/(K_d + c^n)$, where $\Delta\lambda$ is the wavelength shift for each concentration, $\Delta\lambda_{max}$ is



Figure 3.15: (a) Schematic of the surface functionalization and biotin titrations steps: Step A = 150 µg/mL bBSA, Step B = 50 µg/mL SA, Step C = 50 µg/mL BSA, and Step D = biotin dilutions. Each step was washed with a PBS buffer. (b) Complete plots of the small molecule interaction assay, where the blue curve represents the experimental (sensing) data, and the red curve represents the control (reference) data. Light blue portions indicate PSB rinsing steps. (c) Zoomed plots of singlecycle kinetic (SCK) titrations using biotin dilution series at different concentrations. Each concentration was injected at 30 µL/min for 5 min and followed with a 3-min PBS rinse. The grey area shows the 3 σ of the system noise. (d) Wavelength shift as a function of the concentration of biotin. The curve is fitted with the Hill-Langmuir equation with the Hill coefficient of 0.5 and 1, respectively.

the shift corresponding to the saturation of binding sites, c is the analyte concentration, and n is the Hill coefficient. Although it has been reported that there is no cooperativity (n = 1) between the binding sites of the avidin when interacting with biotin [219, 220], an optimized curve fitting (n = 0.5) is applied to the

measurement data with the coefficient of determination R^2 of 0.999 (blue curve in Figure 3.15(d)), indicating a negative binding cooperation. If we assume a maximum binding site occupation happens at the concentration of 10^{-3} M (i.e., $\Delta\lambda_{\text{max}} = 10.97$ pm), the dissociation constant estimated from fitting (n = 0.5) is 1.87×10^{-4} M, larger than the free solution-based affinity value ($K_d = 10^{-15}$ M). Multiple publications have reported the same discrepancy between the binding kinetics of surface-capture vs. solution-phase SA-biotin interactions. This has been theorized to be due to mass transport limitations, altered diffusional characteristics, multivalent binding, and ligand alteration due to the immobilization of SA to a solid substrate [220–222]. By employing the measured $3\sigma_{3row}$ in the fitting equation, a minimum detectable biotin concentration of 2.28×10^{-8} M is obtained.

3.4 Conclusion and discussion

In this chapter, we introduced a novel sub-wavelength grating-based configuration for sensor performance enhancement. The proposed sub-wavelength grating geometry consists of 180×180 nm² silicon pillars periodically in both propagation and transverse directions. The simulated and experimental results of multi-box SWG waveguide indicate that the optical power is largely congregated in the gaps between the Si segments which greatly enhances the overlap between the evanescent field and analyte, thus improving sensor performance.

By employing the multi-box SWG configuration to a microring resonator, bulk and surface sensitivities were characterized and compared with well-studied TMmode-based and conventional SWG-waveguide-based MRR sensors, resulting in 2.5-time and 6-time (TM-mode MRRs), and 1.5-time and 2-time (SWG MRRs) improvement in bulk and surface sensing, respectively. However, there are challenges with this approach. First, although the sensitivity is significantly improved, the bulk iDL of 1.02×10^{-3} RIU and surface iDL of 3.13×10^{-1} nm are comparable to other SWG and other silicon-based sensors as a result of the relatively low measured Q value of 2600 in water. Second, the manufacture of sub-wavelength structures requires either the EBL system, or the most advanced 193 nm immersion lithography technology (demonstrated silicon photonics features down to 50 nm [223]). Several methods can be used to reduce the total losses in multi-box SWG waveguides, such as applying the new advanced fracturing strategies to minimize the scattering losses, single line edge smoothing (SLS), in EBL [224], and employing a Bragg grating resonator instead of the MRR to eliminate the bend radiation losses.

So, we then followed up this research by introducing the phase-shifted Bragg grating resonator in order to reduce the losses of the multi-box SWG-based sensor. Compared to MRR-based counterparts, multi-box PSBG sensors have similar bulk and surface sensitivities but exhibit an improved Q and a smaller footprint (~200 μ m²). Owing to no bend radiation losses, the proposed multi-box PSBG sensor presents higher O around 8000 at 1550 nm wavelengths, leading to an intrinsic DL of 3.6×10^{-4} RIU, which is close to half of that of the well-investigated MRR sensor in the TM mode (7.1 \times 10⁻⁴ RIU). Moreover, because of the highly expanded surface contact area, the multi-box structure is suitable for real-time protein-ligand interaction kinetics monitoring, even for analytes with low MWs. By sequentially introducing biotin dilutions to the streptavidin-modified 3-row PSBG sensor surface, the biotin-SA interaction model is characterized through single-cycle kinetics without regeneration processes. The capability for the detection of biotin dilutions as low as 2.28×10^{-8} M is observed without label attachment, demonstrating the applicability of the proposed multi-box PSBG sensor for detecting small molecules. This may facilitate the development of silicon-based low-cost sensing systems competitive with SPR. However, due to the high noise level (on the order of pm), the experimental result shows a low SNR even in the case of surface saturation. Besides, the immobilized crosslinker on the surface pre-occupies half of the binding sites of the receptor, which cuts down the achievable signal for the analyte binding.

The signal noise from Figure 3.15(c) is mainly generated from the system setup noise (light-source and detector intensity fluctuations or temperature variations) and the biological noise (molecule binding/unbinding). For the biological noise, it can be divided into two parts: intrinsic noise and extrinsic noise. Intrinsic noise is defined as the stochasticity of biochemical interaction of particles, and the extrinsic noise is generated from other non-specific binding processes or from environmental fluctuation, such as molecular machineries, gradients of temperature or chemical concentrations or the coupling of the molecule to the variability of the external environment.

There are several methods that might benefit the improvement of signal quality. First of all, as mentioned in Chapter 1, enlarging the sampling rate (*N*) can effectively increase the SNR with an improvement of \sqrt{N} . Since SNR = μ/σ , where μ is the signal mean or expected value and σ is the standard deviation of the noise. When enlarging the sample number to *N*, the total power of the noise is constant, but σ will reduced by \sqrt{N} . Another method to improve the biological noise floor is improving the affinity between ligand and protein (a smaller dissociation constant). It has been reported that decreasing the dissociation constant can supress the level of random unbinding [225]. In addition, employing another reference as the control group is another widely used method to eliminate the extrinsic biological noise. By subtracting the control group noise level, the noise from the experimental group can eliminate the unnecessary noise generated from the external environment as well as non-specific binding activities.

A small footprint ring resonator sensor is a good approach for sensing in terms of high-multiplexity detection. However, as aforementioned, lower N may cause a lower SNR, thus a worse signal quality. Therefore, a larger sensor size is required for more binding/unbinding interactions synchronously. If we assume the N linearly scales with the area of the sensor surface, the SNR scales with the $\sqrt{(area)}$. It has been demonstrated that sensing devices with larger bio-functionalized surface area have larger SNR [226]. In the aspect of ring resonator-based sensor, because of the resonant-wavelength light trapping mechanism in the resonator, the optical surface area is much larger than the physical surface area of the ring. Therefore, the physical size of the resonator may be not the main restriction to improve the signal quality. Instead, a low concentration of analyte at the sensors surface may degrade the SNR more. Considering a situation that low numbers of analyte molecules are injected to the sensors surface with larger binding affinity. As the surface area increased, the percentage of the surface coverage is reduced, since the number of available molecules is not enough to bind to the available receptors, and hence, the surface area of the device must be optimized. If such optimization is not carried out, the signal will be derived from an area much smaller than the device area, whereas the noise will still be generated from the whole device area, resulting in a degradation of the signal-to-noise ratio.

Chapter 4

Cost-effective Sensing System

4.1 Introduction

A large variety of optical architectures have emerged leveraging the silicon substrate for sensing applications as aforementioned in Chapter 1, however, different signal interrogation configurations are employed based on different architectures. In interferometric sensors, the RI change is converted to an optical phase shift ($\Delta\phi$), thus a phase interrogation is needed; while in resonant sensors, the RI change is converted to a wavelength shift ($\Delta\lambda$), which requires a wavelength interrogation [57]. However, both approaches demand either a wavelength-tunable light source or a high-resolution readout system for precise optical spectrum scanning and processing, which increases the overall cost of the sensing system.

Intensity interrogation schemes have been demonstrated as an alternative solution for sensing, due to their capability for operating the system with a low-cost broadband source input and a relative intensity measurement as the output [227]. Several relevant architectures have been published to low down the system cost [228–230] by using a broadband light source, such as a light-emitting diode (LED) or a superluminescence diode (SLD).

In this section, we introduced two cost-effective broadband light source-based implementations for optical sensing with the intensity interrogation scheme. The first implementation uses a dual-ring sensing system, one of the ring is the RI sensor, and the other ring is a doped silicon photoconductive heater-detector as optical readout. Details can be found in Section 4.2. The second implementation uses the phase-shifted Bragg grating (PSBG) as the sensing component, which is then employed into a symmetric MZI architecture. The RI change in one of the PSBG arms will lead to the phase change at the output of the MZI, thus changing the intensity at the resonant wavelength. Details can be found in Section 4.3.

4.2 Doped silicon-based dual-ring sensing system¹

In an optical biosensing system, the laser is the most expensive building block. Therefore, replacing the laser with a broadband source can provide a lower-cost solution. As Song et al. proposed in Ref. [231], a voltage scanning method for electrical tracking of the changes in the sensing ring with on-chip germanium photodetectors (Ge-PD). This system does not need a high-resolution tunable laser as the input, which not only decreases the cost but also release the detection limit from the laser's resolution (wavelength shifts smaller than the laser linewidth can be tracked). However, to realize the integration of Ge to the SOI substrate, an extra fabrication step, namely the selective area growth by chemical vapour deposition, is commonly required for waveguide photodiodes [232]. Therefore, to reduce the fabrication complexity, we used all-silicon, in-resonator photoconductive heaterdetectors (IRPHDs) to replace the on-chip Ge-PD as the readout in the dual-ring sensing system, which can reduce the overall cost of photonic biosensors, reduce the chip size through decreasing the number of pads by half for electrical I/O, and simplify the control electronics by using a single element to detect light and tune the microring.

4.2.1 Sensing principle

IRPHDs were investigated as a control element in tuning large-scale silicon ring resonator systems [233, 234]. Due to its unique property, which allows the IRPHD to be precisely sensed and tuned simultaneously, the photonic sensor system con-

¹Parts of the section have been published: L. Dias, E. Luan, H. Shoman, H. Jayatilleka, S. Shekhar, L. Chrostowski, and N. A. F. Jaeger, "Cost-effective, CMOS-compatible, label-free biosensors using doped silicon detectors and a broadband source," In CLEO: Applications and Technology, pp. ATu4K-5, *Optical Society of America*, 2019. © by the authors, licensee OSA. Reprinted with permission.

sisting of two rings, a sensor and an IRPHD tracker, can be applied for real-time sensing applications. As the schematic shown in Figure 4.1, the sensing and tracking rings have different radii, which are 8 and 6 μ m, respectively. Rings are made of 90-nm-thick rib waveguide with a core-width of 500 nm. For the tracking ring, the waveguide core was N doped and the sides were N++ doped to form ohmic contacts, presented in Figure 4.1(a). The simulated results of the transmission for the dual-ring system are depicted in Figure 4.1(b). As shown in the grey region, when the sensing and tracking rings are well-aligned in a specific resonant wavelength, the IRPHD photocurrent will be maximized due to the maximized power in the tracking ring. Therefore, the voltage supplied to the IRPHD tracking ring is proportional to the RI-change induced wavelength shift of the sensing ring, when a specific bias voltage is applied to obtain the maximum photocurrent in the IR-PHD. Since the broadband light source has a very large bandwidth compared to FSRs of the sensing and tracking rings, other resonance mode overlaps are possible within the wide bandwidth, which could cause false positives. To ensure only a single resonance mode overlap, we used a 6-nm band-pass tunable filter (BPTF, OTF-950, Santec), which allows the wavelength range containing only one FSR to pass through. The broadband light was then passed through an erbium-doped fiber amplifier (EDFA-C-26G-S, Fiberprime)) to compensate for the BPTF and on-chip grating coupler losses, as the schematic shown in Figure 4.1(c).

4.2.2 Experiments and results

Dual-ring sensing devices were fabricated by IME A*STAR using 193-nm lithography through a Multi Project Wafer (MPW) shuttle run, in Singapore. The total size of the chip is $\sim 3 \times 8 \text{ mm}^2$, which is too small for a PDMS microfluidic chunk bonding on top. Therefore, the chip was encapsulated by the epoxy to form a 2-inch handle wafter. Details about the chip encapsulation process can be found in Chapter 5. A 1-cm-thick PDMS microfluidic chip was permanently bound on the handle wafer, where the sensing ring was aligned to the channel for solution sequencing. A polarization-maintaining optical fiber-array (PLC connections LLC) was used to couple the light from the broadband LED (BeST-SLED, Luxmux) to the chip. Two coplanar GSG probes were employed to detect and tune the tracking ring.



Figure 4.1: (a) Schematic of the IRPHD dual-ring sensing system, including a sensor and a tracker, using a broadband source. (b) Transmission of the dual-ring system across the spectrum of the SLED. The grey area shows the resonance of each ring over a bandwidth of 6 nm, as filtered by the band-pass filter. (c) Block diagram of our experimental setup. (d) Optical microscope image of a test device.

The applied voltage and measured current of the tracking ring were controlled and measured by a source meter (2604B, Keithley). A temperature controller (DC501, Stanford Research System) was used to stabilize the peripheral temperature of the sensing system. The schematic of the sensing setup is presented in Figure 4.2 below.

Before the sensing measurement, optical tests were performed for a dual-ring system, where both rings were IRPHDs and covered with oxide (Figure 4.1(d)), by using the tunable laser (Agilent 81682A, Agilent Technologies) and the power detector (Agilent 81653A, Agilent Technology). As shown in Figure 4.3(a) inset, the test architecture has two grating couplers for the on-chip optical I/O. Therefore, after passing through the BPTF and the EDFA, the guided light was measured at the through port of the tracking ring. The transmission spectrum is depicted in Figure 4.3(a): two peaks are shown within the 6-nm wavelength range, the sensor peak at drop port and the tracker peak at the through port. When two peaks are aligned, most of the power is in the tracking ring and causes the maximum photocurrent. Figure 4.3(b) shows the measured current in the IRPHD ring versus



Figure 4.2: (a) Schematic of setup of the dual-ring sensing system. A fiber array is used to guide the broadband light onto the chip. A microfluidic gasket is used to sequence the solution to the sensing ring. Electrical probes are used to tune and measure the tracking ring. (b) Actual image of the sensing setup.

the applied voltage at different conditions: no optical input (dark current), sensor peak at λ_1 (overlap at λ_1), and at λ_2 (overlap at λ_2). The wavelength shift of the sensor ring was thermally tuned. As compared to the dark current (on the order of milliampere), the photocurrent is quite small (on the order of microampere). Therefore, a calibration step is needed for the photocurrent in the IRPHD by eliminating the dark current, as presented in the inset of Figure 4.3(b).

Sensor performance was characterized by using a series of IPA dilutions from 0% to 20% (v/v), with a flow rate of 30 μ L/min at 25°C. As the sensorgram presented in Figure 4.3(c), voltages for the maximum current are recorded for each IPA concentration by sweeping the applied voltage of the tracking ring, obtaining a bulk sensitivity of ~101.7 V/RIU. To investigate the detection limit of the proposed sensing system, the sensing ring was exposed to DI water for 2.5 hours. The noise floor of the voltage for the maximum photocurrent is depicted in Figure 4.3(d). After detrending, a 3-time standard deviation of 0.1 V was obtained, indicating a system DL of 9.8 $\times 10^{-4}$ RIU.

4.3 Phase-shifted Bragg grating-based interferometric sensing system

In this section, we present a Mach-Zehnder interferometer-based sensing architecture with a cost-effective intensity interrogation scheme, in which two sym-



Figure 4.3: (a) Transmission spectrum of the dual-ring system with the broadband source after the filter and EDFA. Inset: Power is measured from the through port of the tracking ring. (b) Photocurrent as a function of the applied voltage of the IRPHD ring. The blue curve represents the dark current of the IRPHD. The orange and green curves represent the photocurrents when resonant peaks are aligned at λ_1 and λ_2 , respectively. Inset: Zoom-in plot at the bump after the dark current calibration. (c) Sensorgram of the dual-ring system, by introducing IPA dilutions to the sensing ring. (d) System noise of the dual-ring architecture for 2.5 hours. Inset: Noise floor after detrending.

metric arms (sensing and reference arms) consist of a phase-shifted Bragg grating (PSBG). Each PSBG has the same grating parameters. Although similar architectures have been reported recently by combining resonators and an MZI structure for high-performance modulation [235–237], to our knowledge, this is the first experimental demonstration for sensing with an intensity interrogation method. As shown in Figure 4.4(a), when we introduce a solution containing target molecules into the sensing arm, due to the effective index change, the phase at the resonant peak changes at the output, and the constructive or destructive interference occurs when combined with the light from the reference arm. Thus, an intensity interrogation method can be applied to detect the RI change in real-time.

4.3.1 Sensing principle

In a conventional Mach-Zehnder interferometer, the intensity at the output is periodically oscillated based on the phase difference of two arms. In terms of sensing, the light travelling in the sensing arm interacts with the analyte along the distance of L_{sen} , while in the reference arm, light is unexposed to the analyte and can be used for phase comparison [57]. The phase change $(\Delta \phi)$ in the sensing arm can be calculated based on Equation 1.6. As can be seen, a longer sensing arm can amplify the phase change, thus enhance the output intensity variation. To improve sensor performance, interferometer-based biosensors usually have a sensing arm on the order of millimetres or even centimetres, which, however, reduces the onchip sensor density for multiplexable detection. Moreover, because of the high thermal coefficient of silicon, temperature variations can also induce $\Delta \phi$. Increasing the length of the sensing arm also increases the thermal sensitivity of the device and causes a high external thermal noise. Although researchers have proposed several temperature-insensitive interferometer-based devices by introducing different waveguide configurations or propagation modes between two arms, it only supports the temperature independence within a short wavelength range [238, 239].

The combination of resonators and symmetric MZIs can theoretically solve the aforementioned problems. First, in resonator sensors, light-matter interactions are not related to the physical length of the sensing waveguide but depend on the lifetime of a photon staying inside the resonator, which is determined by the quality factor (Q) [14]. Second, using a symmetric MZI structure, in which two arms are exposed in the same cladding material, can remove the temperature effect since two even arms are subjected to the same temperature variation and produce no phase difference. However, it is worth noticing that when the analyte solution is injected into the sensing arm of the resonator-arm-based MZI sensor, not only the phase but the resonance peak will change. This will give a maximum limit for the RI detection when two resonant peaks from sensing and reference arms are totally separated (the sensing peak shifts by one FWHM).



Figure 4.4: (a) Schematic of the PSBG-MZI sensing architecture, showing a phase change (ϕ_{sen}) at the output of the sensing arm due to the analyte attachment and the destructive interference at the MZI output after combining with the light from the reference arm (ϕ_{ref}). (b) Schematic of one PSBG present in both arms. *W* is the waveguide width, ΔW is the corrugation width, Λ is the period, and *N* is the number of periods of the grating on each side.

Among all resonator-based devices, PSBG resonators show great advantages for using in the proposed resonator-arm-based MZI sensor due to its small silicon footprint and high Q (longer optical path lengths). The schematic of a PSBG unit is shown in Figure 4.4(b). W is the width of the core waveguide, ΔW is the corrugation width, and Λ is the period of gratings. The length of the phase-shifted cavity in the middle is $\Lambda/2$, and the grating is replicated N times on each side.

For the proposed PSBG-MZI sensor with the output intensity interrogation, the sensitivity is defined as:

$$S_{\text{PSBG-MZI}} = \frac{\partial I}{\partial \gamma} = \frac{\partial I}{\partial \phi} \times \frac{\partial \phi}{\partial \gamma},$$
 (4.1)

where $\partial \gamma$ is the variation of any physical parameter surrounding the waveguide, $\partial \phi$ is the phase change due to the $\partial \gamma$ in the sensing arm, and ∂I is the output intensity change of the MZI at the resonant wavelength. Similar to the waveguide mode sensitivity (Equation 2.35), $\partial \phi / \partial \gamma$ is waveguide mode dependent, and $\partial I / \partial \phi$ is related to the sensing architecture. In the aspect of MZI-based sensor, output intensity (∂I) is sinusoidal as a function of the $\partial \phi$, but $\partial \phi / \partial \gamma$ is linear. Thus, if we ignore the wavelength shift of the resonant peak and focus on the resonant wavelength only, the output intensity of the PSBG-MZI sensor should follow a sinusodial function of the RI change around the waveguide.

4.3.2 Modelling and simulations

Grating parameters of the PSBG were optimized by using Bloch boundary conditions on one grating periodic cell through a fully vectorized three-dimensional Finite-Difference-Time-Domain (3D-FDTD) approach for band structure calculations, where the periodic cell is repeated infinitely. By adjusting the period (Λ) and corrugation width (ΔW) of the grating cell, we found that when Λ equals 315 nm and ΔW equals 70 nm, the central wavelength of the stop band appears near 1550 nm in the TE mode. Then, we employed the 2.5D variational Finite-Difference-Time-Domain (varFDTD) solver in MODE Solutions to collect simulation results of the PSBG-arm-based MZI. Compared to 3D-FDTD simulations, the 2.5D varFDTD method offers comparable accuracy and versatility but requires less simulation time and memory by replacing the 3D structure with a 2D one. By directly importing the layout of the PSBG-MZI into the 2.5D varFDTD solver, the power transmission and phase response results were collected at the output of the waveguide. All devices were simulated under the water cladding in the TE mode. Figure 4.5(a) shows the simulated transmission of the PSBG-MZI device in the TE mode, where the number of grating periods N is set 100. The phase response at the resonant wavelength is presented in Figure 4.5(b) (red curve), showing a less than $\pi/2$ phase delay compared to the standard waveguide-based MZI (blue curve), which improves RI-based phase modulation efficiency.

4.3.3 Experiments and results

Device fabrication and experimental setup

Our proposed PSBG-MZI devices were designed through an open-source GDS editor KLayout and a publicly accessible Process Design Kit (SiEPIC-EBeam-PDK) [177], and fabricated on silicon-on-insulator (SOI) chips by direct-write 100 keV electron-beam lithography (EBL, JEOL JBX-8100FS) at the University of



Figure 4.5: (a) Simulated power transmission of the PSBG-MZI through 2.5D varFDTD, with the grating parameters Λ of 315 nm and ΔW of 70 nm. The number of periods on each side is fixed (N = 100). (b) Simulated phase response of the PSBG-MZI, as well as a standard waveguid-based MZI with the same length, at the resonant wavelength.

British Columbia and the inductively coupled plasma reactive-ion etching (ICP-RIE) process at Applied Nanotools Inc. As presented in Figure 4.6(a), the PSBG-MZI architecture consists of two symmetric 45- μ m-long PSBG waveguides, where each PSBG has a phase-shifted cavity at the center (Figure 4.6(b)). Two Y-branches are used to split and recombine the guided light through the MZI architecture, the insertion loss of each Y-branch is approximately 3.3 dB (Figure 4.6(c)).

To investigate optical characteristics in the wavelength interrogation, a tunable laser (Agilent 81682A, Agilent Technologies), from 1460 nm to 1580 nm, was used as an optical source. We selected on-chip grating couplers (GC) to couple the light on and off the chip. A polarization-maintaining optical fiber-array (PLC connections LLC) was applied to inject or capture light into or out of devices through these GCs. The output light was then measured with a power detector (Agilent 81653A, Agilent Technology). A temperature controller (LDC501, Stanford Research Systems) was used to control the platform temperature where the chip is placed at. All sensing architectures are aligned towards the center of the chip to facilitate the mounting of poly(dimethylsiloxane) (PDMS) microfluidics on top. One channel was aligned with sensing arms, and the other channel was aligned with reference arms. A syringe pump (Chemyx Nexus 3000, Science Products GmbH) was employed to control flow rates by withdrawing reagents over the sensors.



Figure 4.6: (a) SEM image of the PSBG-MZI architecture, which includes one sensing PSBG-based arm and one reference PSBG-based arm. The length of the two arms is the same. (b) Amplified SEM image of the $\Lambda/2$ phase-shifted cavity in the PSBG. (c) Amplified SEM image of the Y-branch used in the PSBG-MZI.

For the intensity interrogation, we replaced the tunable laser with a cost-effective broadband LED (BeST-SLED, Luxmux) as the optical input. The broadband source was then passed through a tunable optical super-Gaussian filter (OTF-950, Santec) with 3-nm bandwidth at the resonant wavelength, and an erbium-doped fiber amplifier (EDFA-C-26G-S, Fiberprime) to filter out unnecessary wavelengths and compensate for the filter and on-chip GC losses. After passing through the sensing device, the total intensity of the light was detected by the same aforementioned power detector. Schematic of the PSBG-MZI sensing system based on the intensity interrogation scheme is shown in Figure 4.7.

Sensitivity characterizations

The fabricated PSBG-MZI device was first characterized by using the tunable laser, then remeasured by adding a 3-nm bandpass filer. As shown in Figure 4.8(a), the peak power at the resonant wavelength shows a 6.5 dB reduction with the filter, which may include the losses from fiber connections and the filter itself. By replacing the tunable laser with the broadband LED light, we measured the total power intensity at the output of the PSBG-MZI. However, due to the system losses and the filtering processes (both the bandpass filter and the PSBG itself), the measured



Figure 4.7: Schematic of the proposed PSBG-MZI sensing system for the intensity interrogation, which consists of a broadband light source, a tunable filer, an optical amplifier, a sensing device and a photodetector. The RI change caused by analytes is mapped into total intensity variation at the output due to the constructive or destructive interference.

signals at the output is around -75 dBm. Therefore, an optical amplifier was introduced to the system, which enhanced the power intensity out of the PSBG-MZI up to -24 dBm. To evaluate the power consumption in the PSBG-MZI, we injected the light into a calibration device (two GCs with a 150- μ m-long waveguide in between) and received the power intensity of -16 dBm, indicating 8 dB losses or reflection in the PSBG-MZI device. The system noise floor of the proposed PSBG-MZI was evaluated by constantly measuring the output intensity as the water flowing in two channels. After 100 min of measurement, the system shows a noise (3 σ) of 0.22 dB (Figure 4.8(b)).

The temperature stability of the PSBG-MZI device was characterized. Although the symmetric MZI system is theoretically temperature independent, the transmission peak power is not constant with various temperatures, as shown in Figure 4.9. With the temperature increase, resonant peaks of both PSBGs drift (69 pm/°C) and maintain remain in phase, matching our hypothesis. However, the ER of the resonant peak various and reaches the maximum at 29°C, where the critical coupling happens. Therefore, by measuring the output intensity, the proposed PSBG-MZI device shows a thermal intensity change of 0.3 dB/°C, which may cause temperature-based intensity noise.



Figure 4.8: (a) Transmission spectra of the proposed PSBG-MZI sensor with a tunable laser as the input. The blue curve represents the plot with no bandpass filter, and the red curve represents the plot after a 3-nm-bandpass filter. (b) Total intensity detected by the power detector through the PSBG-MZI device as a function of time, showing a system noise (3σ) of 0.22 dB.



Figure 4.9: (a) Resonant peak of the PSBG-MZI device with various temperatures from 23 to 35°C, indicating a thermal shift of 69 pm/°C. (b) Total power intensity of the PSBG-MZI device from 25 to 28°C, indicating a thermal intensity change of 0.3 dB/°C.

Sensor performance was characterized by using a series of IPA solutions and layer-by-layer electrostatic polymer depositions. After calibrating out the losses from the system (including the losses from grating couplers, routing waveguides, fiber connections, etc.), measured sensing results are depicted in Figure 4.10 and Figure 4.11. By injecting IPA dilutions from 1% to 10% (v/v) to the sensing arm sequentially, while keeping ID water flow in the reference arm, the total power intensity detected at the output was measured as a function of time (Figure 4.10(a)), showing an averaged bulk sensitivity of about 810 dB/RIU (Figure 4.10(b)). The temperature was controlled at 25°C during the experiment to eliminate the thermal noise. However, due to the sinusoidal shape of the interferometric spectrum, the phase variation-caused intensity change versus the RI change is not a linear function, which is one of the main drawbacks of MZI-based sensors [39]. Therefore, to evaluate the non-linear sensor response, we introduced a layer-by-layer electrostatic polymer deposition approach (aforementioned in Chapter 3) by continuously injecting polystyrene sulfonate (PSS) and polyallylamine hydrochloride (PAH) to the sensor. After the initial adhesion of positively charged PEI to ensure sufficient coverage, PSS/PAH bilayers were deposited on the sensing arm surface. Each deposition was followed with a 5-min Tris buffer flushing to avoid precipitation and clogging. The sensorgram of 9-PSS/PAH-bilayers deposition is depicted in Figure 4.11, showing a sinusoidal intensity variation at the output, which matches the previous assumption. The strangeness of the first bilayer deposition plot can be explained by the insufficient surface coverage and random protein orientations [175]. If we assume the thickness of the bilayer is constant (1.99 nm), the highest intensity change happens at $\pi/2$ phase difference of two arms, approximately 0.76 dB/nm. Therefore, to obtain the best sensor performance of the proposed PSBG-MZI sensor, an initial $\pi/2$ phase difference is needed, which can be realized by using a heater or injecting a high RI solution to one of the arms. In addition, based on the measured system noise floor of 0.22 dB, the detection limit of the proposed PSBG-MZI sensor is 2.7×10^{-4} RIU and 0.289 nm for bulk and surface sensing, respectively.



Figure 4.10: (a) Calibrated power intensity at the output of the proposed PSBG-MZI sensor with various concentrations of IPA injected as a function of time. (b) Bulk sensitivity of the proposed PSBG-MZI sensor with an averaged value of 810 dB/RIU.



Figure 4.11: Calibrated power intensity in terms of the deposition of polymers at the sensor surface. Yellow and pink areas represent the PSS and PAH injection. Grey areas are the Tris buffer rinse.

4.4 Conclusion and discussion

By employing a broadband light source as the optical input, we developed two costeffective implementations for optical real-time sensing, depending on the intensity interrogation scheme. For the doped silicon-based dual-ring sensing system, the on-chip Ge-PD is no longer a required component as the readout for the sensing system, thanks to the outstanding optoelectronic performance of the IRPHD. Besides, the overall cost, the fabrication complexity, as well as electrical I/O numbers, are further reduced compared to the Ge-PD-based system. However, the drawbacks are also obvious. The dark current of the IRPHD is on the order of milliamps, roughly 1000 times of the sensing signal, which makes it necessary to eliminate the dark current every time to obtain the precise photocurrent. In addition, the IRPHD generates much heat during the measurement, which may influence the resonant peak position of the sensing ring. Therefore, a thermal insulation process (e.t., silicon substrate under-etch) or a safety distance between sensing and IRPHD rings are required. Future work will focus on system optimization, including increasing the responsivity of the IRPHD and reducing the dark current.

For the phase-shifted Bragg grating-based interferometric sensing system, by introducing a PSBG to the MZI device, the optical length of the sensing arm is extended due to the FP cavity, which enhances the light-matter interactions compared to the standard waveguide of the same length. Compared to the MRR, the PSBG offers a smaller footprint, lower intrinsic losses, as well as a better fabrication tolerance. Therefore, PSBG-based devices can have a higher on-chip sensor density and more stable resonant peak positions, which is well suitable for large-scale deep-UV lithography-based fabrications. Furthermore, optical components, such as the bandpass filter, the polarizer, and the photodetector can be integrated on the SOI platform by using advanced CMOS technology. Future direction aims to improve the readout signal, such as using the TM mode input light, which can generate a 3-fold phase change compared to the TE one, and cascading the PSBGs head-to-tail in each MZI arm as the phase amplification.
Chapter 5

System-level Integration Sensors¹

5.1 Introduction

Researchers have continuously broken through sensor performance of photonic sensing devices in the past years. However, system-level integration, which remains the most critical challenge for the technology to be transferred to the clinical world [240], has received less attention. The majority of the research has focused on the enhancement of sensing devices using passive architectures, i.e., all-optical input/output, where microfluidic and optical integrations rely on large dies, from tens to hundreds of square millimeters, connected to smaller microfluidic gaskets in order to keep part of the die exposed to the air for optical fiber coupling [60]. This paradigm remains the same for active architectures, i.e., electrical input and/or output, since electrical probing is also required [241, 242]. Given the small footprint of the sensing devices, most of the die area is only used as mechanical support for the microfluidic system. The advantage of circuit compactness offered by silicon photonics is therefore lost. This not only impacts the unit cost, which is

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proportional to the die area, but also drastically limits the integration of advanced microfluidic systems.

In this section, we developed a system-level architecture based on optoelectronic silicon photonic dies allowing a single optical input to be distributed to multiple MRRs, each one individually monitored in real-time by an on-chip germanium (Ge) photodetector (PD). Details about silicon dies design and manufacturing are presented in Section 5.2. Section 5.3 describes processes of the Fan-Out Wafer-Level-Packaging (FOWLP) we have adapted to a lab-scale process, which consists in encapsulating individual dies into an epoxy-based substrate to create a reconstituted wafer where the dies are further away from each other [243]. Die and package bondpads are interconnected in a fan-out way by one or several metal redistribution layers (RDL). Experimental results about the packaged sensing architectures are shown in Section 5.4.

5.2 Design and fabrication

5.2.1 Die design

The active silicon photonic die was designed using KLayout with the SiEPIC PDK [177] and Pyxis Layout (Mentor Graphics). Simulations for the design of the grating coupler (GC) and MRR were performed with Lumerical FDTD and MODE Solutions, as well as MATLAB (R2017a, MathWorks). As shown in Figure 5.1(a) below, the silicon photonic die is at the center of the packaged biochip, which consists of the silicon photonic die, the fan-out metal trace pattern, and microfluidic channels. In the zoom-in image (Figure 5.1(b)), there are 16 MRR-based sensing devices on the $1 \times 1 \text{ mm}^2$ active die, each of the MRR is connected to one germanium photodetector independently as the electrical output. There is a standard GC in the middle of the die to couple the input light onto the sensing chip.

Germanium photodetectors (Ge-PDs) offer the ability to convert light into an electrical signal directly on-chip, which is a game-changing technology in the architecture of silicon photonic systems [244]. PDs greatly simplify the optical setup since only optical inputs are needed. They also remove the power loss of optical out-coupling, which relaxes the optical power budget. Moreover, their small foot-



Figure 5.1: (a) Layout image of a packaged chip, where the sensing die is in the middle, surrounded by fan-out metal traces and covered by microfluidic channels. (b) Zoom-in image of the layout of the $1 \times 1 \text{ mm}^2$ active die, which consists of 16 MRRs with on-chip Ge-PDs as the electrical readout. (c) Schematic of the cross-section of the proposed packaged chip.

print allows to easily place up to sixteen around the edge of the die (the size of standard bondpads are the limiting factor in that case). PDs are generally optimized for high frequency operation (tens of GHz) for high-speed telecommunication applications. They are used in the photoconductive mode since it makes their response faster [245]. In this mode, they are reverse-biased by a few volts and the photocurrent is amplified by trans-impedance amplifiers (TIA). Their responsivity is measured in A/W and is linear until the photocurrent saturates for optical power of a few tens of milliwatt [246]. For sensing applications, the sampling frequency usually lies below 10 kHz, and the photovoltage between the anode and cathode is measured directly.

5.2.2 Die manufacturing

The dies were fabricated by the Institute of Microelectronics (IME) foundry in Singapore using 193-nm lithography through a Multi Project Wafer (MPW) shuttle run offered by CMC Micro-systems and the SiEPIC program (Canada). $3 \times 8 \text{ mm}^2$ design blocks were thinned down to 250 µm in order to reduce die protrusion from the epoxy package. The blocks were then diced into roughly $1 \times 1 \text{ mm}^2$ dies (including a 50-µm-wide deep trench border) using a dicing saw (DAD3240, Disco). The SEM images of the singulated die are presented in Figure 5.2 below. The oxide cladding above the MRR sensor has been removed through the anisotropic etching process (Figure 5.2(b)). However, in Figure 5.2(c), it is clear to observe that the cladding removal process overetched in the open window area, around 800 nm down to the substrate. From Chapter 2, we know that a substrate overetch sensing architecture may improve sensor performance. For standard waveguide-based architecture, the influence of the overetch is minimized due to the strong optical confinement in the Si core. The effective index of the overetch waveguide is reduced compared to the standard one, therefore, the coupling efficiency of the MRR will change and cause the ER change.

5.3 Lab-scale Fan-Out Wafer-Lavel-Packaging

FOWLP was introduced by Freescale and Infineon in the mid 2000s [247, 248]. This packaging technique consists in encapsulating many individual dies into an Encapsulant Mold Compound (EMC) to create a reconstituted wafer where the dies are further apart from each other. Input/Output interconnects can then be redistributed over the die plus the surrounding ECM (Figure 5.3(a)). It particularly applies to small dies with a high number of electrical connections which cannot be packaged by Fan-In WLP. In contrast to standard Fan-In WLP flows, in FOWLP, the wafer is diced first then precisely positioned on a carrier, maintaining the space for fan-out around each die. The carrier is then reconstituted by molding, followed by making redistribution layer on top of the entire molded area, and forming solder balls on top. FOWLP has now become a well-established technique with first commercialization in the late 2000s. It offers several advantages such as parallel processing of tens or hundreds of dies on 12-inch reconstituted wafers, small



Figure 5.2: (a) SEM image of the singulated $1 \times 1 \text{ mm}^2$ active silicon die. (b) Zoom-in SEM image of sensing MRRs where the oxide cladding has been removed. (c) Zoom-in SEM of the waveguide from the sensor ring. The sample was tilted showing a 800-nm-deep overetch geometry down to the BOX layer.

package footprint, complex multi-redistribution layers, and moreover, it enables the co-integration of dies of different sizes, thicknesses, and technologies. Note that this planar packaging is only compatible with grating couplers, and not with edge couplers, for the optical coupling of silicon photonic circuits. The transition from circular wafers to larger rectangular panels should also allow to reduce the fabrication costs further [249]. The same packaging technique can be used to mass produce low-cost biochips.

Although numerous articles explain this packaging technique, details of the fabrication protocol are generally not published. Moreover, 12-inch reconstituted wafers can only be made with high-end equipment of prohibitive cost for most research laboratories. We have therefore developed a process to fabricate 2-inch reconstituted wafers using very standard cleanroom equipment in order to proto-



Figure 5.3: Schematic of (a) industrial and (b) lab-scale Fan-Out Wafer-Level Packaging processes: i) The silicon dies are placed front side down on a carrier covered with an adhesion layer; ii) A liquid encapsulant mold compound is applied and thermally cured around the dies; iii) The reconstituted wafer is removed from the carrier with sharp edge cut off

, and can be used in standard planar processing; iv) Singulation of packaged dies;

v) The microfluidic layers is integrated on the packaged dies for sensing.

type and test packaged dies before contracting a production facility. As mentioned in Chapter 1, a method to encapsulate single CMOS dies for biosensing applications was recently published in Ref. [166]. As shown in Figure 5.4 below, the authors use a 2-inch Pyrex petridish covered by a 2.5-mm-thick layer of PDMS as a carrier. After placing the die front side down in the middle of the petridish, a liquid epoxy is poured and cured on a hotplate. During the curing, the reconstituted wafer curls away from the petridish due to the thermo-mechanical stress. Such fabrication protocol enables the packaging of CMOS dies but also suffers from several drawbacks which make system integration challenging: the flatness of the epoxy handle wafer is not well controlled, neither is the die-to-epoxy height; the thickness homogeneity requires a perfectly levelled hotplate and thin wafers are difficult to obtain (presented in Figure 5.4). Herein, we propose a protocol (Figure 5.3(b)) closer to FOWLP which makes the fabrication easier, faster, and better controlled,



Figure 5.4: Schematic of a recently published FOWLP process using a petridish covered by PDMS carrier. Due to the thermo-mechanical stress, the fabricated handle wafer is not flatness controlled, and the dieto-epoxy height is over 10 μ m which causes a high risk of discontinuity of the fan-out metal traces from the silicon die to the epoxy substrate.

which also simplifies the system integration. We use 275-µm-thick CMOS dies of $1 \times 1 \text{ mm}^2$ as test vehicles.

5.3.1 Die encapsulation

Our carrier was made of a 3×4 -inch glass slide (Ted Pella Inc.) where PDMS (SYLGARD 184, Dow Corning) with an elastomer base to curing agent ratio of 10:1 was spin-coated for 1 min at 1000 rpm (the mixed uncured PDMS was previously degassed for 2 hours). Just after spin-coating, a 1.8-mm-thick 2-inch internal diameter silicone O-ring (super-resilient high-temperature Silicone, McMaster-Carr) was placed at the center of the glass slide. The PDMS was then cured overnight at 65°C. This led to a PDMS thickness of 30 µm. A liquid epoxy (Durapot 863, Cotronics) with a base to hardener ratio of 10:7 was thoroughly mixed manually for 5 min before being mixed and defoamed using a centrifugal mixer (Thinky) for another 5 min. The reason we selected Durapot 863 as the encapsulant material is that such epoxy is proved to have the required combination of properties, namely, the coefficient of thermal expansion is very close to Si ($2.6 \times 10^{-6}/^{\circ}$ C), which can reduce interfacial stress [166]. The mixed epoxy was then

degassed at 200 mBar for 4 hours (pot life is about 48 hours). A single dummy CMOS die (not the real aforementioned sensing die from Figure 5.2) was manually placed frontside down on the PDMS in the center of the O-ring, gently pressed with tweezers and left for 5 min at 65° C to ensure good adhesion. Cleanliness of the PDMS and die surfaces was crucial at this step. The epoxy was poured in the O-ring region, but not directly onto the backside of the die to avoid trapping air bubbles at the die edge. At room temperature, the epoxy is viscous enough to create a visible dome once it reaches the top of the O-ring. A drop of epoxy was deposited in the middle of another PDMS-coated glass slide (same parameters as for the carrier) which was then flipped upside-down and put in contact with the top of the epoxy dome. The top glass slide was gently lowered to ensure a uniform spreading of the epoxy until it reached the O-ring. Some epoxy may overflow outside of the O-ring but its viscosity contains it within the two glass slides. The sandwich formation was then cured in an oven for 3 days at 110°C and cooled down to room temperature at 1°C/min. Since epoxy does not adhere to silicone, all the elements can be easily separated manually to release the reconstituted wafer. After the O-ring was removed, the side of the wafer is manually pressed to break the thinnest region. The wafer was finally cleaned with acetone, isopropanol and DI water to remove any epoxy residues and debris. The profiles were acquired with a mechanical profilometer (Dektak XT, Bruker).

The important structural properties of the reconstituted wafer are summarized in Table 5.1 and shown in Figure 5.5 below. We experimentally observed that thicker PDMS layers, higher curing temperatures and larger dies tend to increase the die-to-epoxy height difference. For milimeter-thick PDMS layers and with the epoxy cured at 150° C, our test vehicle CMOS die protrudes by almost 14 µm, while 1 cm² dies can protrude by more than 30 µm. Trapping the epoxy between two glass slides significantly improves the flatness of the wafer on both sides and is enough to eliminate the dome shape created by the petridish and by the thermomechanical stress mentioned in Ref. [166]. We found that without the top glass slide the 2-inch wafers exhibit a 500-µm-tall dome shape with a backside having the inverse concavity in addition to higher and longer edge defects. The top glass slide also removes the need for perfectly levelled ovens or hotplates, and enables precise control of the wafer thickness. The edge defect on the frontside is 50-µm-

Structural properties	Performance			
Die-to-epoxy height	$\leq 6 \ \mu m$			
Frontside flatness	\leq 70 µm (\leq 25 µm after edge grinding)			
Backside flatness	$\leq 15 \ \mu m$			
Wafer thickness homogeneity	$\leq 150 \ \mu m \ (\leq 0.2^{\circ} \ frontside-backside \ parallelism)$			

 Table 5.1: Structural properties of 2-inch epoxy reconstituted wafers (lowest-to-highest point)

tall and 200- μ m-long, but could be easily ground out, resulting in a flatness usually better than 25 μ m. The yield of this encapsulation process was found to be about 50%, as some dies exhibit sub-micron infiltration of epoxy which may interfere with the exposed sensing regions (but without necessarily preventing a successful packaging).

Keeping the dies levelled with respect to the epoxy and the reconstituted wafer entirely flat will improve the quality and reproducibility of the following planar processing steps and facilitates the optical and microfluidic integration. Note that multiple dies could also be encapsulated and post-processed on the same wafer if the drift caused by the encapsulation is compensated at the die positioning step [250].

5.3.2 Redistribution layers

The wafer was cleaned with acetone, isopropanol and DI water, dried with a nitrogen gun and dehydrated at 80°C for 5 min. The negative photoresist NR9-3000PY (Futurrex Inc.) was spin-coated for 40 sec at 3000 rpm and soft-baked for 5 min at 120°C, resulting in a 3.5-µm-thick layer. Photolithography was performed with a Mylar film photo-mask (Infinite Graphics Inc) with a minimum feature size of 15 µm. Once the photomask was aligned with the die, the wafer was exposed in a mask aligner (Canon PLA-501F) for 3 min (6 × 30 sec at 1 min intervals, 7.5 mW/cm²) and post-baked for 5 min at 120°C. The wafer was finally developed in RD6 developer for 1 min, rinsed with water and dry with nitrogen. The metal deposition was performed by electron-beam evaporation (Dee Wong 2000) at 5 × 10^{-6} Torr. 10 nm of chromium and 250 nm of gold are evaporated at 0.3 nm/s.



Figure 5.5: Epoxy encapsulation results. (a) A 275- μ m-thick CMOS die of 1 \times 1 mm² encapsulated in epoxy as prototyping. (b) Horizontal profile of the encapsulated die across the lowest lateral pads. (c) Full profile of the epoxy wafer frontside showing the die in the center and a 50- μ m-high and 200- μ m-wide defect at the edge. (d) Full profile of the epoxy wafer backside.

After this first evaporation, the wafer was removed from the evaporator. The wafer was finally immersed in acetone for 10 min to lift-off the resist, and rinsed with isopropanol and DI water.

As shown in Figure 5.6, the quality of the encapsulation permits to perform standard photolithography with a resolution sufficient to easily separate the fan-out traces (20 μ m-wide gap close to the die). Note that the fan-out patterns should be designed slightly larger than the die bondpads to ensure the complete electrical passivation. No cracks were found at the die edge, and electron-beam evaporation led to a good electrical connection (sputtering is also a good alternative). In our case, electrical connection issues rather arise from the die itself which exhibits



Figure 5.6: Results of the redistribution layer patterning. (a) Encapsulated die after fan-out patterning. (b) Optical zoom on the die edge showing the continuity of the fan-out traces. (c) End of the passivation layer just before the fan-out pads. (d) Picture of the reconstituted 2-inch epoxy wafer showing the 12 fan-out traces on a $16 \times 16 \text{ mm}^2$ package.

deep bondpads. Micro probes placed between the fan-out pads and the die bondpads showed a DC resistance of 16 ohm and test structures inside the die could be successfully measured. In a few cases, the SiO_2 passivation generated cracks in the fan-out traces, which may have led to the failure of the electrical continuity. However, the electrical passivation is only needed if the fluid comes in contact with the metal traces. An appropriate design would, therefore, avoid any contact, for instance, by placing all the bondpads on the top and bottom edges of the die, and with the microfluidic channels going across the left and right edges.

5.3.3 Microfluidics layer

The microfluidic channels were patterned using the permanent negative photoresist SU8-3050 (Microchem). The reconstituted wafer was dehydrated at 105° C for 5 min. Once at room temperature, SU8 was spin-coated at 2000 rpm for 30 s, leading to a final thickness of 75 μ m. After being soft-baked 50 min at 105°C, the Mylar photomask was aligned with the redistribution layer and the wafer was exposed during 1 min 30 sec (3 \times 30 sec at 1 min intervals). After a 6-min post-exposure-bake at 105°C, the resist was developed in SU8 developer for 5 min and rinsed with IPA. The wafer was finally exposed without any mask for 1 min and hard-baked at 120°C for 15 min to fully cross-link the resist.

In order to protect the surface of the packaged die during the following singulation step, the positive photoresist AZ4110 (Microchemicals GmbH) was spincoated at 4000 rpm for 40 sec and soft-baked at 70°C for 10 min. Singulation is performed by an Oxford Laser (Serie A, 355 nm solid state laser). The protective photoresist was finally removed with acetone, and rinsed with isopropanol.

Figure 5.7(a) shows the patterning of two 150- μ m-wide micro-channels crossing the die. Note that the photomasks have air bubbles trapped inside the Mylar film which create small defects on the SU8 surface after photolithography. Each channel is terminated by 1-mm-wide reservoirs to relax the alignment tolerance of a fluidic manifold. As many as 8 different micro-channels could be integrated on the 16 × 16 mm² package (Figure 5.7(b)). As shown in Figure 5.7(c), the SU8 layer almost completely hides the topography of the die, which results in a flatness better than 10 μ m at the top of the resist layer. This enables an easy integration into a pressure-sealed fluidic manifold. Figure 5.7(d) shows the final packaged die after singulation. One thousand of these packages could be fabricated at a time on a single 24-inch × 18-inch reconstituted panel [249].

The whole FOWLP packaging process (from a single $1 \times 1 \text{ mm}^2$ silicon die to the $16 \times 16 \text{ mm}^2$ packaged chip) for the designed 16-MRR-PD-based silicon photonic sensing die (as shown in Figure 5.1(b)) is presented in Figure 5.8 below, which does not involve any plasma treatment since it could damage nonelectrostatic discharge (ESD) protected electronic devices.

Our protocol accurately mimics FOWLP but only relies on simple consumables and standard cleanroom equipment. The sensor die protrusion is small enough to enable reliable electrical interconnects and is almost completely hidden by the SU8 layer (Figure 5.9(b)). As shown in Figure 5.8 after singulation from the reconstituted wafer, the $16 \times 16 \text{ mm}^2$ package is large enough to be easily handled. Each



Figure 5.7: Results of the microfluidic layer patterning. (a) Packaged die with two 150- μ m-wide micro-channels. (b) Top left corner of a packaged die with height 50- μ m-wide micro-channels (no passivation layer on the gold). (c) Profile of the package perpendicular to the channels (Y axis) and along the part between the two channels (X axis). (d) Picture of the 16 × 16 mm² packaged CMOS die after singulation.

micro-channel is connected to 1-mm-wide inlet and outlet ports, which strongly relax the fluidic gasket alignment and complexity to mechanically seal the open micro-channels. The package bondpads are also 1-mm-wide and are easily interfaced with spring-loaded pins for electrical read-out.

The possible toxicity due to leaching epoxy was assessed by immersing an autoclaved packaged die in a culture medium containing MCF7 cells. After 3 days, the cells were able to form a monolayer on the epoxy with a density similar to the one observed on the polystyrene petridish. The minimum size of the silicon die depends less on the photonic sensing devices footprint than on the pitch of the biofunctionalization area that can be made by high-throughput non-contact printers, typically about 200 μ m [251]. 1 × 1 mm² dies can, therefore, be used for up to 8



Figure 5.8: Schematic of the fabrication flow of the $1 \times 1 \text{ mm}^2$ packaged chip by the lab-scale FOWLP process by only using the standard clean-room equipment, which includes the die encapsulation, photolithography, metallization, microfluidics, and singulation steps.

differently functionalized sensors. The first goal of our lab-scale packaging technique is to demonstrate that silicon photonic dies as small as $1 \times 1 \text{ mm}^2$ can be used for biosensing, which significantly reduces the unit cost (a 12-inch wafer has a total area of 70,600 mm²). The second objective is to de-risk commercial translation of the technology by prototyping package dies that accurately resemble the ones which foundries and packaging companies can mass produce. Finally, it also provides designers with a practical means to reduce the R&D costs by decreasing the die area. It can also help the biosensing and telecommunication communities to better manage the design space.



Figure 5.9: (a) Photograph of the microfluidic channels at the epoxy/die interface, inlet/outlet port and metal redistribution layers at the package and die bondpads. (b) Profile of the package before and after the SU8 microfluidic layer patterning. The topography of the die corresponds to a photodetector row.

5.4 Experiments and results

5.4.1 Biochips instrumentation

A 6-mm-thick transparent Poly(methyl methacrylate) (PMMA) block was laser cut and drilled to make a fluidic gasket (VLS4.60, Universal Laser Systems). A 0.5-mm-thick poly-dimethylsiloxane layer (PDMS, SYLGARD 184, Dow Corning) was added underneath the PMMA block to improve the sealing of the microchannels. Inlet and outlet holes were punched using needle tips. The biochip set on a copper base thermally controlled by a temperature controller (LDC501, Stanford Research Systems) as shown in Figure 5.10. A syringe pump (Nexus 3000, Chemyx) was used to sequence the reagents through the micro-channels. The biochips were electrically probed using spring-loaded pin arrays (1.27-mm-pitch, 854-22-012-10-001101, MillMax) connected to standard header pins via printed circuit boards (Pcbway) (Figure 5.10). Each header pin was connected to an RC filter with a cut-off frequency of 10 kHz in order to remove high-frequency noise. Digital multimeters (34401A, Agilent) able to internally store 200 measurements in a row were used as analog-to-digital converters (ADCs). A Python script was used to synchronize the laser wavelength sweeps at 0.5 nm/s and the electrical measurements with steps of 20 pm over 4 nm. The sweep interval was set at 18 s. The laser wavelength noise was measured at less than 1 pm using a C-Band Wavelength Calibrator Acetylene Gas Cell (Wavelength References). As depicted in Figure 5.12(a), several sweeps were stitched together. For the bulk sensitivity and bioassay experiments, MATLAB was used to fit the spectra with a smoothing-spline function using a smoothing parameter of 0.9999 to obtain a resolution of 0.1 pm and reduce the noise of the measurements (no interpolation in Figure 5.12(a), and no smoothing in Figure 5.12(b)). The resonant peaks were tracked using a MATLAB-based program developed in-house. A long-working distance fiber focuser (LPF-D4-1550-9/125-S-1-17.3-18AS, OZ Optics Ltd.) was used to project the image of the single-mode fiber (SMF) core onto the GC through the fluidic gasket and microfluidic SU8 layer (Working distance = ~17 mm) depicted in Figure 5.10. The fiber focuser was roughly pre-aligned visually using a visible light source. The position and polarization were finely adjusted before the experiment to optimize the photovoltages with the IR laser.

5.4.2 Photodetector characterization

We characterized a 50 μ m-long PD with a 125- μ m-long waveguide taper. The PD was connected to two grating couplers (GC) through a Y-branch in order to find the best optical alignment and calculate the insertion loss of the GC (Figure 5.11(a)). The polarization was Transverse Electric (TE) and the wavelength was set at λ = 1550 nm. The insertion loss of the Y-branch is estimated at -3.3 dB. We used an Agilent 8164A laser mainframe containing a tunable laser source (Agilent 81682A) with an output wavelength range from 1460-1580 nm and optical power sensors (Agilent N7744A). A fiber array (PLC Connections, 127 μ m-pitch) was aligned on the two GCs, with the laser input on the bottom one. The laser power was then swept from 9 to -9 dBm. Lower optical powers were accessed by moving the input fiber array away from the two GCs to increase the effective insertion loss. For each optical power, 200 measurements were made with a Keithley 2602 (Tektronix Inc.) and were fitted to a Gaussian distribution to determine the mean value and the noise (defined as 3 times σ).

As shown in Figure 5.11, in the photovoltaic regime, the responsivity of the



Figure 5.10: Schematic of the biochip setup installation. The packaged chip is connected to the fluidic gasket and spring-loaded pins. The chip is passively aligned using a shallow corner for abutment milled into a copper base. The gasket consists of a transparent PMMA bock and a PDMS layer which seals the microchannels defined on the package by a SU8 layer. The copper base is thermally controlled by a thermoelectric cooler. The spring-loaded pins are connected via printed circuit boards to headers for flexible prototyping of the readout electronic circuit. A fiber focuser is used to support the optical accessibility of the silicon photonic die through the thick fluidic gasket.

PDs is very non-linear and starts to strongly saturate toward 255 mV for optical power greater than -10 dBm (0.1 mW). Defining a responsivity in V/W would not be very relevant. The logarithmic response can indeed be seen as an advantage since it allows to measured signal over a large range of optical powers. For the OFF state (no light), we measured a residual voltage of 3.5 mV, which leads to maximum ON-to-OFF ratio of 70. The photovoltage noise can be in a first approximation attributed to the photon noise since it decreases as the optical power increases. A signal-to-noise ratio (SNR) better than 1000 can be achieved for an optical power greater than -20 dBm (10 μ W). The sensitivity, defined as the derivative of Figure 5.11(b), is plotted in inset of Figure 5.11(c). It reaches a maximum of 8 mV/dB for an optical power around -25 dBm. Taking into account the trade-off between noise and sensitivity, the optimal operating optical power lies between -10 and -25 dBm. This value dictates the power budget for the laser power and



Figure 5.11: (a) Photo-electrical characteristics of on-chip germanium photodetectors in the photovoltaic regime. (b) Photovoltage response with optical power in dBm scale, and noise in inset. (c) Photovoltage response with optical power in linear scale, and sensitivity in inset. (d) Resonant peak of a TE microring resonator measured by the PD (red line) and by the mainframe (black markers).

insertion loss of the single input grating coupler. The same PDs are expected to exhibit similar or better performances in Transverse Magnetic (TM) polarization and in the O-band ($\lambda = \sim 1310$ nm), as they do in the photodiode regime.

We also demonstrated the ability of the photovoltaic mode to measure resonant peaks, such as the ones from microring resonators used in biosensing. The fiber array was first aligned using the GC number 1 and 2, as depicted by the scheme in the inset of Figure 5.11(d). The laser power was adjusted to read an off-resonance detected power of -14 dBm and the spectrum of a TE ring resonator with a quality factor of 6850 was acquired. The laser was then swapped to the fiber aligned with the GC number 3 and the optical power was adjusted to obtain a photovoltage of 205 mV (-14 dBm received by the PD). The wavelength sweep was finally repeated

while measuring the photovoltage. Figure 5.11(d) plots the two measurements which look very similar due to the almost linear response of the PD in the dB scale. Even at low-light levels, a small amplification is enough for the photovoltage to be read by inexpensive micro-controllers integrating ADCs.

5.4.3 Free space optical coupling and power budget

Efficient optical coupling in silicon photonics remains a critical challenge for industrial applications. Contrary to the telecom sector, optical fibers cannot be permanently attached to disposable biochips due to packaging costs. In the active architecture proposed here, only long-working distance optical coupling is possible since the input beam has to go through the entire fluidic gasket. When using grating couplers originally optimized for SMFs, the non-contact operation is made possible by using a fiber focuser, as in optical scanners [252]. Note that the inplane alignment tolerance (X and Y axes) is the same as for SMFs, here measured at $\pm 2.5 \,\mu\text{m}$ for a 1 dB power penalty. Fiber focusers also exhibit a longitudinal alignment tolerance (Z axis), here measured at $\pm 20 \ \mu$ m. In contrast with optical scanners, here, the optical alignment is only required once. With our fiber focuser, and without gasket, an insertion loss (IL_{GC}) of 7 dB can be obtained on bare silicon dies at a working distance of 17 mm. Optical aberrations due to the fluidic gasket induce an additional loss which increases as a function of the incident angle and the gasket thickness. This excess loss has been measured at 0.9 dB for a GC with an incident angle of 10° and a 5-mm-thick gasket. In our design, the incident angle is 20° and the gasket is 1 mm thicker, therefore, we expect IL_{GC} to reach an optimum of $\sim 12 \text{ dB}$.

Other sources of loss should also be considered to calculate the power budget. Each Y-branch adds ~0.3 dB loss in addition to the 50% splitting leading to $IL_{Y-branch} = \sim 3.3$ dB. For the TM mode, the waveguide propagation loss L_{prop} is usually close to 2 dB/cm. Note that only a very short section of bus waveguides are exposed to the aqueous medium via the annular oxide opening, so water absorption loss can be neglected. Bend loss can also be neglected for bend radii larger than 20 µm [53]. The required laser power P_{laser} can be calculated using the following equation:

$$\frac{P_{\text{laser}}}{P_{\text{PD}}} = IL_{\text{GC}} + log_2(N_{\text{PD}}) \times IL_{\text{Y-branch}} + L_{\text{wg}} \times L_{\text{prop}},$$
(5.1)

where N_{PD} is the number of PDs (assumed to be a power of 2), L_{wg} is the length of the waveguide, and P_{PD} is the optical power required at each PD. Here, the power ratio for 16 MRR is 25.4 dB. Given a required P_{PD} of -30 dBm, we obtain $P_{\text{laser}} = -4.6$ dBm (0.35 mW). Biochips with more PDs, hence more sensors, are therefore within the reach of most tunable lasers.

5.4.4 Sensitivity characterization and real-time biosensing

Figure 5.12(a) shows the spectrum of a single MRR in pure water from 1480 to 1560 nm wavelengths. It should be mentioned that the photovoltage could be increased by designing a GC with a smaller incident angle. Figure 5.12(b) offers a closer look at 6 MRRs which exhibit an extinction ratio of 7 dB and a quality factor of 4000. For this MRR design, the extinction ratio and quality factor can be better than 25 dB and 10,000, respectively [18]. Lower values are a result of the MRR being non-critically coupled which can be attributed to the oxide overetch presented in Figure 5.2 since the overetch waveguIde geometry reduces the effective index and changes the waveguide propagation mode, which results in an imperfect coupling. For the bulk sensitivity assessment, the sensorgram shown in Figure 5.12(c) leads to $S_{\text{bulk}} = 220 \text{ nm/RIU}$ (Figure 5.12(d)), as expected by simulation. Finally, the biosensing performance of the active biochip was evaluated by a standard sandwich bioassay to probe biomolecular interactions with specific and non-specific targets, as depicted in Figure 5.12(e). The final wavelength shift shown in the sensorgram of Figure 5.12(f) is in good agreement with the one obtained with a passive architecture using the same MRR design [18]. The system noise level (3σ) calculated over the entire bioassay is measured at 11.3 pm and 11.5 pm for the two MRRs respectively (Figure 5.12(f), inset), leading to a bulk DL of 5.2×10^{-5} RIU. These noise levels are in good agreement with the ones reported on passive architectures in Ref. [25]. Sensor performance could be improved by employing higher quality factor resonant cavities (Bragg grating, or micro-disk resonators) or sub-wavelength grating waveguides with better bulk and surface sensitivities discussed in Chapter 2, and Chapter 3.



Figure 5.12: (a) Spectrum of a MRR over a wide wavelength range. (b) Overlapped spectra of 6 MRRs over a wavelength range slightly larger than one free spectral range. (c) Sensorgram of bulk refractive index steps of NaCl dilutions in pure water. (d) Measurement of the bulk sensitivity at 220 nm/RIU by linear regression. (e) Schematic of the protein assay in the sensing channel. (f) Sensorgram of the bioassay.

5.5 Conclusion and discussion

We have advanced the state-of-the-art silicon pho-tonic biosensors by designing a novel system architecture based on active optical-in/electrical-out circuits. Using Fan-Out Wafer-Level-Packaging, we could reduce the silicon die size while increasing the effective surface for microfluidic integration and electrical interconnects. The disposable biochips can be entirely manufactured by the most recent, yet mature, low-cost mass production technologies, leading to competitive fabrication costs of highly integrated Lab-on-Package biochips. The active architecture also improves the compactness, cost and robustness of the reader device since it enables the use of slower tunable lasers, simpler and more tolerant fluidic gaskets, simpler optical alignment (eventually passive) and easier biochip handling by non-experts. To the best of our knowledge, this is the first demonstration of a fan-out packaged silicon photonic system. In the short-term, fan-out packaged active silicon photonic biosensors will allow the integration of actuators, such as electric field-driven microfluidics for sample manipulation and preparation [253], and other sensors, such as CMOS-compatible electrochemical biosensors. In the mid-term, on-chip lasers could be seamlessly integrated, thus taking full advantage of all-electrical integration and optical biosensing.

While the supply chain already exists for the fan-out packaged biochips, some system aspects would need to be improved in order to transfer this approach to the clinical domain. First, the optical coupling needs to be more robust to mechanical vibrations and misalignment. This can be achieved by using a fiber collimator and giant grating couplers, for which 1-dB power penalty can be up to \pm 50 μm for lateral misalignment and in the milimeter range for axial misalignment [254]. A collimated beam would also suffer less from optical aberration due to the fluidic gasket. Such grating couplers require a very weak grating which, in CMOS foundries, can be obtained by a custom shallow etch or by 193-nm immersion lithography [223]. Ultimately, passive alignment could be sufficient, thus allowing plug-and-play style photonic biochips. Secondly, since fast optical scanning is not required, the bench-top high-end laser source could be replaced by off-theshelf compact and affordable DBR or DFB lasers for which the wavelength can be tuned either using temperature or current control. This solution has recently been demonstrated in a hand-held device [255]. For the read-out electronics, digital multimeters can be replaced by commercial ADC chips integrated on a custom PCB, along with the spring-loaded pins, low-pass filters and a micro-controller. This could also reduce the noise level of the photodetectors. Finally, the fluidic gasket needs to allow easy replacements into the reader device. It could rely on a low-cost molded plastic cartridge in which the fan-out packaged biochip would be integrated, as recently developed in Ref [256].

Chapter 6

Conclusion and Future Work

6.1 Summary

In Summary, we have theoretically and experimentally studied silicon photonicbased resonator sensors and sensing systems for performance enhancement and overall cost reduction, respectively. Two novel sub-wavelength grating-based waveguide configurations have been designed, simulated, and characterized to improve the sensitivity of the sensors. Then, by applying the low-cost broadband light source with the intensity interrogation scheme, two types of cost-efficient silicon photonic sensing systems were proposed and characterized. Besides, a systemlevel integration leveraging the Fan-Out Wafer-Level-Packaging has been developed to achieve low-cost sensor packaging. The major contributions of this dissertation include:

- Design and demonstration of high sensitivity substrate-overetch SWG microring resonator-based sensing devices. By employing the anisotropic etching process, the SWG waveguide achieves ~300 nm overetch in the BOX layer, with an optimized etch selectivity SiO₂/Si of 10:1. The SOE-SWG configuration shows an enhanced bulk sensitivity up to 575 nm/RIU, roughly 1.2-time of the standard SWG waveguide sensor, and a waveguide sensitivity over one.
- Design and demonstration of high sensitivity multi-box SWG sensing de-

vices. The bulk sensitivity obtained from the multi-box is ~580 nm/RIU, and the surface sensitivity is around 1900 pm/nm, 2-fold of the standard SWG-based sensor. By replacing the MRR architecture with a phase-shifted Bragg grating, the Q is 2-time improved, which enables a more stable noise floor and a lower detection limit. Furthermore, the capability of the proposed multi-box SWG PSBG sensor for the detection of small molecule interactions is also demonstrated, indicating a minimum detectable biotin concentration down to 2.3×10^{-8} M.

- Design and demonstration of two cost-efficient SOI-based sensing systems. The first system consists of two cascaded MRRs, one of the rings exposed to analyte solutions serves as the sensing ring, and the other ring which is an all-silicon, in-resonator photoconductive heater-detector (IRPHD) serves as the monitor. The other system is made of a symmetric MZI, in which both arms consist of the PSBG. Both systems enable the low-cost broadband light source as the input and use the optical intensity interrogation scheme for readout. In addition, both systems are CMOS compatible, allowing optical components integrations, such as the filter, the polarizer, and the photodetector, in the same SOI platform.
- Design and demonstration of the MRR-based sensing system-level-integrated architecture with active optical-in, electrical-out circuits, which shows similar sensor performance compared to the passive counterpart. By developing the lab-scale Fan-Out Wafer-Level Packaging process, which only relies on standard cleanroom equipment, we successfully reduced the silicon active die size while increasing the effective surface for microfluidic and electrical integrations, which paves the way for the deployment of the cost-efficient silicon photonic-based sensing system.

6.2 Conclusion

6.2.1 High sensor performance devices

Throughout the research described in Chapter 2 and Chapter 3, it can be concluded that using SWG metamaterial-based waveguide configurations can enhance the sensitivity due to the reduced optical confinement in the waveguide core, which can be utilized as next-generation high-sensitivity silicon photonic sensing devices. Table 6.1 compares the performance of our proposed SWG-based sensing architectures in this dissertation with several advanced SWG waveguide-based sensors demonstrated on the same 220-nm SOI platform. All of them present the bulk sensitivity around 400 to 600 nm/RIU but as the sacrifice of the quality factor (lower than 10k), which results in a relatively low detection limit. Compared to MRRs, other sensing architectures such as Bragg gratings and photonic crystals show a better *Q* due to no bend radiation losses. Among them, multi-box SWG waveguide-based Bragg grating sensing architect shows the best detection limit (3.6×10^{-6} RIU), which is closer to the theoretical iDL in aqueous solutions at 1550 nm (2.4×10^{-6} RIU).

As compared to the bulk sensitivity, the surface sensitivity is of greater importance. To enhance the surface sensing performance, the most direct thing is to increase the intensity of the light at the surface of the waveguide, and the light is highly confined at the surface only, which can eliminate the bulk sensing signal. In addition, the molecule of interest should be localized in the region where light has the strongest intensity. Such as Ref. [257] demonstrated, the distribution of molecule of interest will influence sensors performance in a slot waveguide-based sensor. Therefore, in order to enhance the sensitivity, a same-size microfluidics should be finely aligned to the slot region, where the molecules can only be distributed within the slot region. Last, the overall noise floor is another important factor that can influence the fundamental limit. Beyond the set-up noise, the biological noise is another major source of noise in the sensing system. To eliminate the non-specific binding caused by the interferent, a pre-biofunctionalization at the sensors surface is required, which has strong specific interaction with the molecule of interest. Moreover, the sensors surface should be large enough to have a decent

Sensor type	Geometry	Q (k)	S_{bulk} (RIU ⁻¹)	$S_{\text{surf}} (\text{nm}^{-1})$	iDL (RIU)
MZI	Slot	N/A	598 nm	N/A	N/A [182]
PHC	Multi-slot	4.2	586 nm	N/A	$6.29 imes 10^{-4}$ [258]
MRR	Multi-box	2.6	580 nm	2.05 nm	$1.03 imes 10^{-3}$
PSBG	Multi-box	7.8	580 nm	1.91 nm	$3.6 imes 10^{-4}$
MRR	SOE	1.5	575 nm	N/A	$1.75 \text{ nm} \times 10^{-3}$
MRR	Pedestal	<1.8	545 nm	2.30 nm	$> 1.58 imes 10^{-3}$ [179]
MRR	TM-mode	9.8	429 nm	N/A	$3.71 imes 10^{-4}$ [84]
Theoretical S _{bulk} in aqueous solutions at 1550 nm wavelengths					1165 nm/RIU [36]
Theoretical iDL in aqueous solutions at 1550 nm wavelengths					2.4×10^{-4} RIU [44]

 Table 6.1: Sensor performance comparisons of the state-of-the-art SWGbased architectures.

number (N) of binding and unbinding interactions happening at the same time in order to maintain a good signal-to-noise-ratio, which gives \sqrt{N} improvement [36].

A trade-off between the sensitivity and the quality factor is observed in resonatorbased sensors: to achieve a high *S*, the optical mode should be distributed more into the target analytes, whereas, to maintain a high *Q*, the confinement of the optical mode propagating in the waveguide has to be strong for low losses [48]. In addition to the inherent losses associated with the sensor architecture, external factors such as temperature fluctuations, misalignment of the I/O fiber due to vibration, as well as the light source intensity fluctuations can also create noise and affect the detection limit of the sensing system [44]. It is worth to notice that for a low *Q* device, the dominant noise source is typically amplitude noise, however, for a high *Q* device, the temperature fluctuations become the dominant factor [43]. The relationship between the *S* and the *Q* of published silicon photonic-based sensing devices is presented in Figure 6.1. Compared to other architectures, sub-wavelength gratings-based geometry enhance the sensitivity effectively, which speeds up the development of high-performance optical sensors, but further investigations and optimizations are still required to reduce the overall losses.

Although SWG metamaterials show appealing performance in the application of label-free detection and the capability for the detection of small molecular weight analytes, such as pesticides and cannabis, there is still "a long way to go" before commercialization. Until now, SWG-based configurations can only be re-



Figure 6.1: Experimental data points of published sensing architectures on the SOI platform. For the light travelling exclusively in water, the intrinsic Q is 1.29×10^5 for critical coupling.

alized by the EBL process due to the strict requirement of the silicon size (on the order of tens of nanometers), which restricts the low-cost mass production through DUV processes. Advanced Micro Foundry (AMF), in Singapore, provides the MPW run with the minimum feature size down to 120 nm, which gives the possibility to realize SWG geometries by CMOS fabs. However, transmission spectra are still not acceptable, due to the high side-wall roughness and the overexposure on silicon blocks, as compared in Figure 6.2, which strongly increase the overall losses in SWG waveguides. Recently developed wet immersion lithography technique gives the potential for the transfer of the SWG fabrication with satisfactory performance to CMOS fabs with a silicon features down to 50 nm [223].

6.2.2 Cost-effective active sensing systems

The investigation of the sensing system compatible with broadband light sources and intensity interrogation schemes, as discussed in Chapter 4, relieves the budget for purchasing the expensive tunable laser, and the high-resolution photode-



Figure 6.2: (a) SEM image of the SWG based MRR by deep-UV litho processes, and Zoom-in SEM images of SWG pillars by deep-UV litho processes. The pink rectangle represents the as-draw silicon block. (b) Same SWG configurations fabricated by EBL processes. The pink rectangle represents the as-draw silicon blocks.

tector or the optical spectrum analyzer. Leveraging the CMOS fabs, active components, such as on-chip lasers, LEDs, semiconductor optical amplifiers (SOA), tunable bandpass filters, and photodetectors, can be realized in the same SOI platform through direct mounting, heterogeneous or monolithic integration methods. Since each wafer can hold hundreds of active sensing dies, this will result in rapid declines in cost down to several dollars per die, which will enable the unit cost of the sensor chip to be effectively controlled even with the on-chip light source.

Thanks to the FOWLP technique (in Chapter 5), the size of the active silicon die can be scaled down as small as millimeter square while maintaining the sensing complexity and accuracy, as well as allowing the on-chip microfluidic system and signal transfer circuits, which offers an economic approach for low-cost sensor system wafer-level integration suitable for commercialization. The key factor defining possibility of the sensor that can be applied for local diagnostics is the dimension of the sensor. Compared to the bulky size of the apparatus of SPR systems, a benchtop and user-friendly instrument is required to avoid costly and inconvenient core facilities, which matches well with the development trend of silicon photonic sensing architectures. Although, only simple silicon photonic waveguide-based sensing devices can be fabricated by the DUV lithography with relatively low performance (50-200 nm/RIU), commercial products based on the CMOS fabs have been realized by Genalyte, Inc. [259] and SiDx, Inc. [260] for chemistry, hematology, infectious disease, and urinalysis detection with advantages of low-cost, small footprint, high-throughput and rapid response. Some present-day commercial products based

on the evanescent field sensing technologies are listed in Table 6.2. Taking advantages of the CMOS fabs, cost-effective silicon photonic-based sensing devices and systems enriched the modern market for applications in local diagnostics and personalized medicine.

In conclusion, in this dissertation, we presented several novel sensing architectures for the application of the low-cost and high-sensitivity real-time monitoring. By leveraging different fabrication foundries, EBL or deep-UV lithography, silicon photonic-based sensors with different waveguide geometries can be selected for the specific biosensing problems. Such as by applying the multi-box or the substrate over-etched SWG geometries, targets of interest such as pesticides, cannabis and other small molecules can be detected, which can be used for the development of hand-held cannabis detection kits or the development of long-term water quality monitoring devices. Although current deep-UD lithography technique may not allow such small feature size, further development of the lithography schemes, such as immersion lithography, will solve the problem. In terms of the broadband source-based architecture, to consist the goal of low-cost sensing, CMOS foundries-compatible waveguide geometries, such as the strip and slot waveguides, were employed. Although these sensors do not present advanced sensor performance, the reduced fabrication cost with production amount, as well as the capability of light source and readout components integration, gives these sensing devices the potential to be used for fast and low-cost home-care IVD detection, including the blood typing and pregnancy tests. However, silicon photonic-based sensing is still under development, a plenty of issues are still needed to be solved. Such as the biological noise caused by the non-specific binding and system set-up noise are of great importance to be solved in order to compete with the gold-standard technique, SPR systems. In addition, the optimization of surface bio-functionalization is also important, since the concentration of targets of interest in real samples is usually quite small compared to the interferents. A strong, robust and renewable receptor layer is high desired. Therefore, there are still aspects that can be developed, improved and optimized in the study of silicon photonic-based sensors.

Technology	Manufacturer	Instrument	Integration	Reference
SPR	GE Healthcare	Biacore-T100	4 channels	Ref. [261]
	Nyicoya	OpenSPR	2 channels	Ref. [262]
Optical gratings	Axela	DotLab	1 channel	Ref. [39]
Resonant gratings	Corning	EPIC System	384-well plate	Ref. [263]
TE-MRR	Genalyte	Maverick	128 rings	Ref. [259]
TM-MRR	SiDx	N/A	\geq 64 rings	Ref. [260]

 Table 6.2: Some commercialized examples of present-day evanescent fieldbased optical sensing.

6.3 Future work

With the accomplished works for optimizing sensor performance and reducing the overall cost in this dissertation, suggested improvements and future work include:

- A more thorough study of noises involved in the SWG-based sensor, both from the device and environmental variations, will be investigated. The overall losses of the SWG sensor need to be reduced in order to achieve a detection limit on the order of 10⁻⁴ RIU. Strategies such as using the single line edge smoothing (SLS) technique [224] or replacing the fabrication material with Si₃N₄ to reduce the scattering loss, manufacturing on the wafer with a 300 or 500 nm silicon layer thickness to reduce the substrate leakage loss, and decreasing the operational wavelength to 1310 nm for a lower water absorption are well-worth of investigation.
- Light sources integration and connection through the PWB technique. Costefficient and technically viable integration of light sources onto PICs is one of the key challenges of integrated optics. Recently, Billah et al. demonstrated that photonic wire bonds could address the challenge by fabricating 3D freeform waveguides between multi-chips [264]. It is worth for the investigation of the light source integration to the proposed FOWLP chips (multichip integration) as a step towards the lab-on-a-chip medical diagnostics system, as presented in Figure 6.3. In addition, electronic components for signal processing can also be integrated onto the chip through GlobalFoundries CMOS techniques [265], such as transimpedance amplifiers (TIA), analog-



Figure 6.3: (a) 3D schematic of the light source integration onto the reconstituted wafer by the PWB technique. Both the sensing and the laser chips are packaged on the same substrate, and the electrical interconnects are patterned by FOWLP processes. (b) Cross-section image of the proposed integration. The 3D freeform waveguide is covered by a protecting layer. The distance between the two chips should be less than 30 μ m. (c) Schematic of a lab-on-a-chip sensing chip, including the reconstituted sensor chip and electronic circuits for signal processing, both of which can be realized through CMOS techniques.

to-digital converters (ADC) and digital-to-analog converters (DAC), and receivers, which further facilitates the development of on-chip sensing systems.

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Appendix A

List of Publications

A.1 Journal publications

- Stephen Lin, Mustafa Hammood, Han Yun, Enxiao Luan, Nicolas A. F. Jaeger, and Lukas Chrostowski. "Computational lithography for silicon photonics design." IEEE Journal of Selected Topics in Quantum Electronics (2019).
- Enxiao Luan, Kashif M. Awan, Karen C. Cheung, and Lukas Chrostowski. "High performance sub-wavelength grating-based resonator sensors with substrate overetch." Optics Letters 44, no. 24 (2019): 5981-5984.
- Enxiao Luan, Han Yun, Minglei Ma, Daniel M. Ratner, Karen C. Cheung, and Lukas Chrostowski. "Label-free biosensing with a multi-box subwavelength phase-shifted Bragg grating waveguide." Biomedical Optics Express 10, no. 9 (2019): 4825-4838.
- Lukas Chrostowski, Hossam Shoman, Mustafa Hammood, Han Yun, Jaspreet Jhoja, Enxiao Luan, Stephen Lin et al. "Silicon Photonic Circuit Design Using Rapid Prototyping Foundry Process Design Kits." IEEE Journal of Selected Topics in Quantum Electronics (2019).
- 5. Loic Laplatine*, **Enxiao Luan***, Karen C. Cheung, Daniel M. Ratner, Yonathan Dattner, and Lukas Chrostowski. "System-level integration of active sili-

con photonic biosensors using Fan-Out Wafer-Level-Packaging for low cost and multiplexed point-of-care diagnostic testing." Sensors and Actuators B: Chemical 273 (2018): 1610-1617.

- Enxiao Luan, Hossam Shoman, Daniel M. Ratner, Karen C. Cheung, and Lukas Chrostowski. "Silicon photonic biosensors using label-free detection." Sensors 18, no. 10 (2018): 3519.
- Enxiao Luan, Han Yun, Loic Laplatine, Yonathan Dattner, Daniel M. Ratner, Karen C. Cheung, and Lukas Chrostowski. "Enhanced sensitivity of subwavelength multibox waveguide microring resonator label-free biosensors." IEEE Journal of Selected Topics in Quantum Electronics 25, no. 3 (2018): 1-11.
- Shane Patrick, Richard J. Bojko, Stefan JH Stammberger, Enxiao Luan, and Lukas Chrostowski. "Improvement of silicon waveguide transmission by advanced e-beam lithography data fracturing strategies." Journal of Vacuum Science & Technology B, Nanotechnology and Microelectronics: Materials, Processing, Measurement, and Phenomena 35, no. 6 (2017): 06G504.

A.2 Conference proceedings

- Enxiao Luan, Valentina Donzella, Karen C. Cheung, and Lukas Chrostowski. "Advances in Silicon Photonic Sensors Using Sub-Wavelength Gratings." In 2019 24th OptoElectronics and Communications Conference (OECC) and 2019 International Conference on Photonics in Switching and Computing (PSC), pp. 1-3. IEEE, 2019.
- Donald Witt, Enxiao Luan, Kashif Masud Awan, Jaspreet Jhoja, and Lukas Chrostowski. "Teaching undergraduate students integrated photonics and fabrication through research." In Education and Training in Optics and Photonics, p. 11143_150. Optical Society of America, 2019.
- 3. Leanne Dias, **Enxiao Luan**, Hossam Shoman, Hasitha Jayatilleka, Sudip Shekhar, Lukas Chrostowski, and Nicolas AF Jaeger. "Cost-effective, CMOS-

compatible, label-free biosensors using doped silicon detectors and a broadband source." In CLEO: Applications and Technology, pp. ATu4K-5. Optical Society of America, 2019

- Enxiao Luan, Loic Laplatine, Jonas Flueckiger, Osama Al'Mrayat, Daniel M. Ratner, Karen C. Cheung, and Lukas Chrostowski. "System-Level Integrated Active Silicon Photonic Biosensor for Detecting Small Molecule Interactions." In 2018 IEEE 15th International Conference on Group IV Photonics (GFP), pp. 1-2. IEEE, 2018.
- Enxiao Luan, Han Yun, Loic Laplatine, Jonas Flueckiger, Yonathan Dattner, Daniel M. Ratner, Karen C. Cheung, and Lukas Chrostowski. "Subwavelength multi-box waveguide-based label-free sensors." In Integrated Optics: Devices, Materials, and Technologies XXII, vol. 10535, p. 105350H. International Society for Optics and Photonics, 2018.
- Loic Laplatine, Osama Al'Mrayat, Enxiao Luan, Carter Fang, Shayan Rezaiezaden, Daniel M. Ratner, Karen C. Cheung, Yonathan Dattner, and Lukas Chrostowski. "System-level integration of active silicon photonic biosensors." In Microfluidics, BioMEMS, and Medical Microsystems XV, vol. 10061, p. 100610I. International Society for Optics and Photonics, 2017.

Appendix B

Derivation of Figures of Merit of Silicon Photonic Sensors

In this part, we compare sensor performance of two principal silicon photonic optical sensing configurations, i.e. Mach-Zehnder interferometer (MZI) and microring resonator (MRR). Using the perturbation analysis [56], there figures of merit, such as the sensitivity (S) and the detection limit (DL), are derived and compared.

B.1 Sensitivity derivation

B.1.1 Resonator-based sensors

In a resonator, wavelengths which are resonantly supported will be trapped inside the resonator, thus presenting resonant peaks in transmission plots. In an MRRbased sensor, the total ring is usually exposed to the analyte solution. When we introduce a refractive index perturbation factor q into the sensing system, the resonant condition will change, and result in a resonant wavelength shift.

Based on the resonant condition, light that is coupled into the resonator travels a total circle and presents a phase reproduction at the beginning point, as presented in Equation B.1:

$$\phi_1 = \beta_1 L = 2\pi n \quad (n \in Z^+) \tag{B.1}$$

where β_1 is the propagation constant in the resonator, and L is its circumference.

After introducing a change q, the refractive index around the sensor changes; and the propagation constant is changed as well ($\delta\beta_q$). Thus, a new resonant peak will appear at different wavelengths (λ). Because of dispersion (refractive indices of Si and SiO₂ are wavelength dependent) a change in the propagation constant due to change in the resonance's wavelength occurs ($\delta\beta_\lambda$), which returns the propagation constant back to its original value. So the total propagation constant after introducing a change q will be $\beta_2 = \beta_1 + \delta\beta_q + \delta\beta_\lambda$ ($\beta_1 \ge \beta_2$). Thus, the new phase shift after a total circle at the new resonant wavelength is:

$$\phi_2 = \beta_2 L = (\beta_1 + \delta \beta_q + \delta \beta_\lambda) L = 2\pi n. \tag{B.2}$$

Due to $\phi_1 = \phi_2 = 2\pi n$, we find out that $\beta_1 = \beta_2 = \beta_1 + \delta \beta_q + \delta \beta_\lambda$. Thus, $\delta \beta_q + \delta \beta_\lambda = 0$. For $\delta \beta_q$, it follows:

$$\delta\beta_{\rm q} = \frac{2\pi\delta n_{\rm eff}^{\rm q}}{\lambda}.\tag{B.3}$$

where $\delta n_{\text{eff}}^{\text{q}}$ is the effective index change caused by the q factor. Thus,

$$\delta\beta_{\lambda} = -\frac{2\pi\delta n_{\rm eff}^{\rm q}}{\lambda}.\tag{B.4}$$

Then, we make a derivative of λ , showing (β decreases as λ increases):

$$\frac{\Delta\beta_{\lambda}}{\Delta\lambda} = -\frac{\partial\beta}{\partial\lambda} = -\frac{2\pi}{\lambda^2} \left(\lambda \frac{\partial n_{\rm eff}^{\rm q}}{\partial\lambda} - n_{\rm eff}^{\rm q}\right),\tag{B.5}$$

where $\lambda \frac{\partial n_{\text{eff}}^{q}}{\partial \lambda} - n_{\text{eff}}^{q} = -n_{\text{g}}$. Thus, from Equation B.4 we put $\partial \beta_{\lambda}$ as β change ($\partial \beta_{\lambda} \leq 0$, thus $\Delta \beta_{\lambda} = -\partial \beta_{\lambda}$) caused by the different λ into the upper equation:

$$\Delta \lambda = \Delta \beta_{\lambda} \frac{\lambda^2}{2\pi n_{\rm g}} = \frac{2\pi \Delta n_{\rm eff}^{\rm q}}{\lambda} \frac{\lambda^2}{2\pi n_{\rm g}}$$
(B.6)

After simplification, we get:

$$\Delta \lambda = \frac{\lambda \Delta n_{\rm eff}^{\rm q}}{n_{\rm g}}.$$
 (B.7)

For sensing purpose, it is usually defined as the wavelength shift ($\Delta\lambda$) due to the

refractive index change caused by a q factor (Δn_{add}^{q}), with a unit of [nm/RIU]. Thus the sensitivity of a MRR-based senor is:

$$S_{\rm MRR} = \frac{\Delta\lambda}{\Delta n_{\rm add}^{\rm q}} = \frac{\lambda}{n_{\rm g}} \left(\frac{\partial n_{\rm eff}^{\rm q}}{\partial n_{\rm add}^{\rm q}}\right),\tag{B.8}$$

which also indicates that the sensitivity of a microring resonator is not related to its size.

B.1.2 Interferometer-based sensors

As for the MZI-based sensor, the sensing area is usually one of its arms with the length of L_1 , and the other arm serves as the reference covered with oxide (L_2). Due to the phase difference between the sensing arm and the reference arm, constructive or destructive interference will happen at the output.

By changing the refractive index above the sensing arm, the interference plot will shift. For the sensing purpose, we usually track the destructive point (extremum point with π phase difference) for the phase shifting, where:

$$\beta_1 L_1 - \beta_2 L_2 = (2n - 1)\pi \ (n \in Z^+) \tag{B.9}$$

After introducing the q to the sensing arm, the phase difference at the new wavelength λ will be:

$$(\beta_1 + \delta\beta_1^{\mathbf{q}} + \delta\beta_1^{\lambda})L_1 - (\beta_2 + \delta\beta_2^{\lambda})L_2 = (2n-1)\pi$$
(B.10)

Thus, from Equation B.9 and B.10 we get:

$$\delta\beta_1^q L_1 + \delta\beta_1^\lambda L_1 - \delta\beta_2^\lambda L_2 = 0. \tag{B.11}$$

For $\delta\beta_1^q$ and $\delta\beta_1^\lambda$ from the sensing arm, we know from Equation B.3 and B.5 that:

$$\Delta \beta_1^{\rm q} = \frac{2\pi \Delta n_{\rm eff}^{\rm q}}{\lambda} \tag{B.12}$$

and

$$\Delta\beta_1^{\lambda} = \Delta\lambda \frac{2\pi}{\lambda^2} n_{\rm g1} \tag{B.13}$$

where n_{g1} is the group index of the sensing arm. As for the reference arm, the propagation constant is similar (n_{g2} is the group index in the reference arm):

$$\Delta \beta_2^{\lambda} = \Delta \lambda \frac{2\pi}{\lambda^2} n_{\rm g2}. \tag{B.14}$$

Since $\delta\beta_1^{\lambda}$ and $\delta\beta_2^{\lambda}$ are negative values, $\Delta\beta_1^{\lambda} = -\delta\beta_1^{\lambda}$ and $\Delta\beta_2^{\lambda} = -\delta\beta_2^{\lambda}$. Therefore, by bringing Equation B.12, B.13 and B.14 into Equation B.11, we achieve:

$$\frac{2\pi}{\lambda}\Delta n_{\rm eff}^{\rm q}L_1 - \Delta\lambda \frac{2\pi}{\lambda^2} n_{\rm g1}L_1 + \Delta\lambda \frac{2\pi}{\lambda^2} n_{\rm g2}L_2 = 0.$$
 (B.15)

After simplification, we get:

$$\frac{\Delta\lambda}{\Delta n_{\rm eff}^{\rm q}} = \frac{\lambda L_1}{n_{\rm g1}L_1 - n_{\rm g2}L_2} \tag{B.16}$$

which indicates that the sensitivity of a MZI sensor scales with $\frac{L_1}{L_1-L_2}$ ($n_{g1} \approx n_{g2}$ even after a *q* introduced effective index change). Therefore, the sensitivity of a MZI-based sensor is:

$$S_{\text{MZI}} = \frac{\Delta\lambda}{\Delta n_{\text{add}}^{\text{q}}} = \frac{\lambda L_1}{n_{\text{g1}}L_1 - n_{\text{g2}}L_2} (\frac{\partial n_{\text{eff}}^{\text{q}}}{\partial n_{\text{add}}^{\text{q}}}) \approx \frac{L_1}{\Delta L} \frac{\lambda}{n_{\text{g}}} (\frac{\partial n_{\text{eff}}^{\text{q}}}{\partial n_{\text{add}}^{\text{q}}}).$$
(B.17)

In another situation, the MZI is treated as a unity, and both arms are exposed to the same solution ($\Delta L = L_1 - L_2 \neq 0$). Therefore, the wavelength shift due to the cladding RI change is no longer caused by the phase difference between two arms. The transmission spectrum moves as a whole because of the change of β in both arms, which is similar to the thermal drift of Si devices. Under this condition, the MZI has the same sensitivity formula (Equation B.8) and is independent of its size.

B.2 Detection limit derivation

Another important factor in evaluating the performance of a sensor is the detection limit (DL). As described in Chapter 1, the detection limit of a RI sensor is specified

as the minimum RI change necessary for the detectable signal at the output, which is related to the sensitivity and the system noise (Equation 1.5). However, due to the limitation of its definition, the detection limit (or the system detection limit) is not an objective parameter for the comparison between sensors. Therefore, the intrinsic detection limit is introduced. Derivations of two types of DLs, i.e., the system detection limit (sDL) and the intrinsic detection limit (iDL), are presented below.

B.2.1 Intrinsic detection limit of resonator-based sensors

The iDL was introduced as a substitute for resonant sensors, which is only dependent on intrinsic characteristics, as expressed in Equation 1.10. From Equation 1.9, we know $Q = \frac{\lambda}{\Delta \lambda_{\text{FWHM}}}$. Thus, the iDL equals $\Delta \lambda_{\text{FWHM}}/S$.

The full width at half maximum (FWHM) indicates the optical wavelength width of the resonant peak at which the transmitted intensity is equal to half (-3 dB) of its maximum value. In a all-pass ring resonator, the transmission function is:

$$T = \frac{a^2 - 2arcos\phi + t^2}{1 - 2atcos\phi + (at)^2}$$
(B.18)

where *a* is the single-pass amplitude transmission $(a^2 = exp(-\alpha L), \alpha$ is the power attenuation with [1/cm]), and *t* is the self-coupling coefficient. When this resonator is at resonant wavelengths, from Equations B.1 we know $\phi = 2\pi n$, thus $T = (a - t)^2/(1 - at)^2$. So for critical coupling where a = t, T = 0 and no coupled light comes out of the resonator. While, for the FWHM in the resonator, T = 1/2, thus from Equation B.18 we get:

$$\cos\phi = \frac{2a^2 + 2t^2 - (at)^2 - 1}{2at}.$$
 (B.19)

For a small ϕ , we have:

$$\cos\phi = 1 - \frac{\phi^2}{2}.\tag{B.20}$$

After simplification, we have:

$$\phi^2 = \frac{(1-at)^2 - 2(a-t)^2}{at}.$$
(B.21)

Since a = t, we get $\phi = \sqrt{\frac{(1-at)^2}{at}}$. According to Equation B.1 and B.5, we get following equations after a derivative of λ :

$$\frac{\Delta\phi}{\Delta\lambda} = \frac{\Delta\beta}{\Delta\lambda}L = -\frac{2\pi L(\lambda\partial\frac{n_{\rm eff}}{\partial\lambda} - n_{\rm eff})}{\lambda^2} = \frac{2\pi Ln_g}{\lambda^2}.$$
 (B.22)

So $\Delta \lambda = \Delta \phi \lambda^2 / 2\pi n_g L$. For the FWHM, $\Delta \lambda_{FWHM} = 2\Delta \lambda$, thus we have (where $\Delta \phi = (1 - at) / \sqrt{at}$) [13]:

$$\Delta\lambda_{\rm FWHM} = \frac{(1-at)\lambda^2}{\pi n_{\rm g}L\sqrt{at}}$$
(B.23)

Therefore, the iDL of a MRR sensor is:

$$iDL_{\rm MRR} = \frac{\Delta\lambda_{\rm FWHM}}{S} = \frac{(1-at)\lambda}{\sqrt{at}\pi L} \frac{\partial n_{\rm add}}{\partial n_{\rm eff}}.$$
 (B.24)

B.2.2 Intrinsic detection limit of interferometer-based sensors

However, for MZI-based sensors, no such metrics are proposed. That is due to the sinusoidal shape of the interferometric spectrum, which fixes the linewidth of the FWHM to be half of the FSR and is independent of the loss. Hence, in a MZI-based sensor, if we derive in the same way, the iDL is only related to its sensitivity, i.e., to the length of its sensing arm (L_1) :

$$iDL_{\rm MZI} = \frac{\lambda}{2L_1} \left(\frac{\partial n_{\rm add}}{\partial n_{\rm eff}} \right).$$
 (B.25)