An optofluidic metasurface for lateral flow-through detection of breast cancer biomarker

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A B S T R A C T

The rapid growth of point-of-care tests demands for biosensors with high sensitivity and small size. This paper demonstrates an optofluidic metasurface that combines silicon-on-insulator (SOI) nanophotonics and nonfluidics to realize a high-performance, lateral flow-through biosensor. The metasurface is made of a periodic array of silicon nanoposts on an SOI substrate, and functionalized with specific receptor molecules. Bonding of a polydimethylsiloxane slab directly onto the surface results in an ultracompact biosensor, where analyte solutions are restricted to flow only in the space between the nanoposts. No flow exists above the nanoposts. This sensor design overcomes the issue with diffusion-limited detection of many other biosensors. The lateral flow-through feature, in conjunction with high-Q resonance modes associated with optical bound states of the metasurface, offers an improved sensitivity to subtle molecule-bonding induced changes in refractive index. The device exhibits a resonance mode around 1550 nm wavelength and provides an index sensitivity of 720 nm/RIU. Biosensing is conducted to detect the epidermal growth factor receptor 2 (ErbB2), a protein biomarker for early-stage breast cancer screening, by monitoring resonance wavelength shifts in response to specific analyte-ligand binding events at the metasurface. The limit of detection of the device is 0.7 ng mL−1 for ErbB2.

1. Introduction

Optical label-free biosensors can detect biomolecules based on their intrinsic physical properties, such as Raman scattering, refractive index, and second harmonic generation (Celebrano et al., 2015; Choi et al., 2010b; Salafsky, 2006; Wu et al., 2012). In particular, many refractive index-based biosensors have been implemented to study analyte-ligand interactions without using labels (Im et al., 2014; Liu et al., 2014; Zhang et al., 2008). In contrast to binding assays that require fluorescent or enzymatic tags, label-free assays often eliminate the need for time-consuming labeling processes and can monitor binding kinetics in real time (Ali et al., 2015; Ali et al., 2017b; Díaz-Diestra et al., 2017). Therefore, label-free biosensors are gaining increasing attention in the fields of life sciences, pharmaceutics, and clinic diagnosis (Bergwer, 2001; Navратilova and Hopkins, 2011; Rich and Myska, 2003; Sridharamurthy et al., 2007; Thillaivinayagalingam et al., 2010; Xue et al., 2014a, 2014b). Recently, optical resonators using surface plasmon resonance, photonic crystal, and whispering gallery mode (Aldorn et al., 2003; Cunningham et al., 2004; Fan et al., 2008; Heeres et al., 2009; Sun and Fan, 2011; Vollmer and Arnold, 2008) have been extensively studied and exploited for label-free biosensors. Owing to their strong ability to confine resonating fields, these biosensors are sensitive to the presence of biomaterials immobilized in the close vicinity of their surfaces (Sun et al., 2016b).

While significant progress has been made to develop optical label-free biosensors, how to efficiently deliver samples to the sensor surface remains challenging. To address the issue with the mass transfer limitation, microfluidic systems have been developed and applied to label-free biosensors (Choi and Cunningham, 2006, 2007). To improve the integration between the sensing and fluidic elements, several optofluidic biosensors (e.g., liquid-core ring resonators (White et al., 2006) and anti-resonant reflecting optical waveguides (Yin et al., 2004)) have been used to facilitate the transport of analyte. Both the microfluidic and optofluidic approaches rely on a flow-over scheme, where liquid samples flow through a channel, during which the analytes diffuse from the sample stream onto the surface of the biosensor (Brolo et al., 2004; Lindquist et al., 2009; Sinton et al., 2008; Wang et al., 2014). Recently, a vertical flow-through sensor design was implemented to label-free optical biosensing, where liquid samples flow through a horizontally...
placed, nanopatterned dielectric or metallic diaphragm with nanoholes that functions as both the sensing element and conduits (Cetin et al., 2014; Eftekhari et al., 2009; Escobedo et al., 2010; Yanik et al., 2010). This sensor has enhanced the interaction between the sensing surface and analytes, thus reducing detection time.

This paper reports on a novel lateral flow-through biosensor, consisting of a metasurface with a two-dimensional (2D) periodic array of silicon nanoposts (SNPs), for the detection of cancer biomarker. The structure is manufactured in the thin top silicon layer of a silicon-on-insulator (SOI) substrate, coated with graphene oxide (GO) nanosheets, and biofunctionalized with specific antibody molecules. A polydimethylsiloxane (PDMS) slab with an inlet and an outlet is bonded to the top surface of the SNPs, thus restricting the flow of liquid analytes in between the PDMS and the buried oxide layer of the SOI substrate (Fig. 1). It is worth noting that silicon or SOI-based metasurfaces have attracted increasing attention due to the flexibility in tuning of their optical properties (Sun et al., 2016a; Zhu et al., 2015, 2013a), and the fabrication compatibility with complementary metal-oxide-semiconductor (CMOS) process. The high refractive index of silicon is favourable for light modulation (Cheung et al., 2012; Fang et al., 2016; Ferrara et al., 2015; Taillaert et al., 2006; Van Laere et al., 2007), e.g., to enhance optical fields. Our biosensor is featured with the lateral flow-through design for improved analyte-ligand interactions at the metasurface. Owing to a reduced diffusion length, the biosensor design will overcome the issue of mass transfer limit that occurs in many existing label-free biosensors (Choi et al., 2010a). Therefore, the biosensor will have an improved sensitivity and a reduced assay time. In addition, the metasurface supports different optical resonance modes, such as the bound states in the continuum (BIC) mode, and leaky waveguide mode (Chang-Hannai and Yang, 2012; Wang et al., 2016a, 2016b), to detect biomolecule absorptions. In this work, the guided mode resonance (GMR) mode, whose linewidth depends on the coupling angle, is utilized and exhibits a high sensitivity to a change in refractive index at the surface of SNPs. The sensor design emphasizes both analyte delivery and sensitivity. The key figure of merit of the device and its ability to detect cancer biomarkers are demonstrated.

2. Experimental section

2.1. Fabrication of the optofluidic metasurface

An SOI wafer is used to fabricate the SNPs. First, 200 nm-thick PMMA is coated onto the substrate at 2000 rpm for 45 s. Subsequently, e-beam lithography is used to pattern the nanoholes array in the PMMA. Next, a 15 nm-thick Al2O3 layer is evaporated using electron-beam evaporation, and then patterned using lift-off process, to form a protection layer during the following deep reactive-ion etching of Si (Fig. 1(b)). After the SNPs are formed, the Al2O3 layer is removed via wet chemical etching. The overall size of the device is 1 \times 1 \text{ mm}^2. To enable laterally flowing liquid analytes through the SNP area, a 2 mm-thick PDMS slab with the pre-drilled inlet and outlet is bonded directly onto the top surface of the SNPs via oxygen plasma treatment. Fig. 1(c) shows the formed nanofluidic channels embedded with the SNPs.

2.2. Setup for optical reflection measurement

A tunable laser (ANDO, AQ4321) is used as a light source providing a wavelength range from 1520 nm and 1620 nm with a central wavelength of 1570 nm. The light is collimated and incident onto the metasurface through a 50/50 beam splitter cube. The biosensor is mounted on a rotation stage to adjust the angle of incidence. The reflection spectrum is measured in real time using an InGaAs photodetector synchronized through an oscilloscope.

2.3. ErbB2 detection assay

The biosensor is used to quantify a well-established breast cancer biomarker, ErbB2 (Ali et al., 2017a). The biofunctionalization of the surface begins with introducing an intermediate layer of GO to the surface. The GO layer allows enhancing the loading capability of anti-ErbB2 molecules. In this step, the metasurface is treated with oxygen plasma for 50 s to make the SNPs hydrophobic. Next, a well-dispersed solution of single-layer GO nanosheets (0.4 mg/mL) is prepared in DI water, followed by thorough sonication for 1 h. 50 \mu L of this solution is drop-cast onto the metasurface and then dried at room temperature (25 °C) for 2 h. 20 \mu L of PBS (pH = 7.4) solution containing anti-ErbB2 molecules (0.24 \mu M) is drop-cast onto the GO-coated metasurface, followed by treating a mixed solution of EDC (1-ethyl-3-(3-dimethylaminopropyl) carbodiimide hydrochloride, 0.2 M) and NHS (N-hydroxysuccinimide, 0.05 M) at a 1:1 ratio (Ali et al., 2016a). The abundant oxygenated groups such as –COOH and –CHO at GO are activated and utilized to make covalent binding with anti-ErbB2 using the EDC-NHS coupling chemistry (Ali et al., 2016b, 2017a). To immobilize antibody molecules, the metasurface is kept in a humidity chamber for 12 h, and then is washed with PBS to remove unbound antibody molecules. The resulting primary amine groups present at anti-ErbB2 bind with carboxyl groups at GO to form strong CH-NH amide bonds. Finally, 2.0 mg/mL of bovine serum albumin molecules is used to block non-specific sites of anti-ErbB2 on the metasurface.

3. Results and discussion

3.1. Nanophotonic and nanofluidic simulations

The SNP array (350 nm thickness) is designed on the top of a 3 \mu m-thick oxide layer to provide an optical resonance around 1550 nm wavelength. The silicon device layer is transparent around this wavelength and has a negligible extinction coefficient \( k < 0.001 \) and a large refractive index \( n = 3.477 \). The geometric parameters of the SNPs include the array period \( A \), nanopost width \( w \), device layer thickness
(\(t_d\)), and duty cycle (\(\varphi = w/\Lambda\)). The basic principle of the SNPs and corresponding optical characteristics have been discussed in our previous work (Wang et al., 2016b). The SNPs exhibit a BIC mode and a GMR mode, depending on the angle of incidence. As illustrated in Fig. 1(a), the angle of incidence is defined in terms of a standard spherical coordinate system, with the polar angle \(\theta_i\) being measured from the \(z\)-axis. Because of the symmetry, the optical responses of the device are polarization-independent when \(\theta_i = 0\).

Rigorous coupled wave analysis (RCWA) is used to study optical resonances and evaluate their performances for detecting biomolecules (Fig. 2). The details of the RCWA simulation method are described in Supplementary information. Fig. 2(a) presents the obtained photonic band diagram of the SNPs obtained by plotting the calculated reflection spectra as a function of \(\theta_i\). The wavelength and the incident angle change from 1450 nm to 1650 nm, and from \(-15\) to \(15^\circ\), respectively. The region indicated by the black box (Fig. 2(a)) contains a BIC mode, which appears at \(\lambda_r = 1536\) nm and \(\theta_i = 0\). For the angles near the BIC mode, the strong reflection regions accommodate high-sensitivity resonance features. Fig. 2(b) compares the simulated reflection spectra for \(\theta_i = 0^\circ, 1^\circ, 2^\circ, 3^\circ, 4^\circ\), and \(5^\circ\). The obtained resonant linewidth decreases as the incidence angle approaches \(0^\circ\). For example, the resonance linewidth at \(\theta_i = 4^\circ\) is found to be 30 nm, corresponding to a Q-factor of 52. In comparison, the Q-factor significantly increases to 900 when \(\theta_i\) is reduced to \(1^\circ\). The vanishing of the linewidth at \(\theta_i = 0^\circ\) implies the existence of BIC.

The near-field distributions of the resonance mode in the \(xz\)-plane (Fig. 1) are calculated using the RCWA method (Fig. 2(c)). The distributions of the \(E_x, E_z\), and \(H_y\) components when \(\lambda_r = 1536\) nm and \(\theta_i = 1^\circ\) are shown. The colour scale represents the amplitudes of the electric and magnetic fields normalized to the amplitude of the incident field. As seen from the field distributions, the maximum field enhancement factor is as high as 300. The enhancement factor of the field intensity is calculated by averaging the electric field intensities within the sensor area (\(-\Lambda/2 < x < \Lambda/2\) and \(0 < z < 350\) nm). As a result, the averaged enhancement factor of the field strength (intensity or amplitude) is \(\sim 2.7 \times 10^5\). The distributions of the tangential components \(E_x\) and \(H_y\) appear asymmetric. The mode can be excited because of the asymmetric nature of the incident wave at \(\theta_i = 1^\circ\). Fig. 3(a) shows the normalized near-field distribution of \(E_x\) in the \(xy\)-plane at the center of SNPs (\(z = 175\) nm). The
immersed in DI water and solutions of ethanol-water mixtures. The spectra were measured at wavelength as a function of the refractive index.

\[ \theta \] increasing 2°, 3°, and 4°. The black curve represents the resonance at \( \theta_i = 1° \). When the BIC mode resides, there is no signature of resonance in the spectrum. When \( \theta_i \) increases, the optical resonances appear as the narrowband dips in the reflected spectra. At \( \theta_i = 1° \), the BIC mode turns into a radiative resonance at \( \lambda_r = 1534 \text{ nm} \) with a linewidth of \( \delta \lambda_r = 5.8 \text{ nm} \). Further increasing \( \theta_i \) from 1° to 4° results in a significant increase of the resonance linewidth, while the resonance wavelength remains near \( \lambda_r = 1537.2 \text{ nm} \). The measured spectra agree well with the simulated results (Fig. 2(b)). To illustrate how to tune the resonance strength by changing \( \theta_i \), the Q-factor is plotted as a function of \( \theta_i \) (Fig. 4(c)). The Q-factor decreases exponentially from 270 to 57 with increasing \( \theta_i \) from 1° to 4°. It is worth noting that the resonance features a flat angular dispersion, which can be exploited to realize refractive index-based sensing with a focused excitation.

3.2.2. Refractometric sensing

Fig. 4(d) compares the measured reflection spectra when the metasurface is covered with different index solutions. The solutions are prepared by mixing deionized (DI) water and ethanol at different ratios of 1:2, 4:5, 1:1, 4:3, 2:1, 4:1, and 1:0, to produce the refractive indices of \( n = 1.340, 1.336, 1.334, 1.332, 1.329, 1.325, \) and 1.318, respectively. The black curve in Fig. 4(d) shows the case when the channel is filled with DI water (\( n = 1.318 \)). The resonance dips are found to shift toward longer wavelengths with increasing the value of \( n \). These reflection spectra are measured, all at \( \theta_i = 1° \). To calculate the index sensitivity (\( S_n \)), the resonance wavelength of the device is plotted as a function of surrounding refractive index (Fig. 4(e)). The resonance wavelengths are found by fitting the data in the vicinity of minimal reflection using a 2nd order polynomial function. The slope of the linear fit in Fig. 4(e) is calculated as the index sensitivity \( S_n = \Delta \lambda_r / \Delta n = 720 \text{ nm/RIU} \).

3.3. Detection of breast cancer biomarker

The SNP-based biosensor can be utilized to detect biomolecules, such as DNA, protein, or small molecules, in real time. As an example, the device is used to quantify a well-established breast cancer biomarker, ErbB2. As a label-free technology, the SNP-based biosensor measures the refractive index change caused by the immobilization of
the ErbB2 molecules. The target ErbB2 molecules are captured by the anti-ErbB2 antibodies on the sensor surface. The details of this label-free assay are described in the Experimental section. Fig. 5(a) summarizes the major assay steps, including subsequent deposition of GO, anti-ErbB2 antibody, and blocker, and the detection of ErbB2 antigen. The SEM image in Fig. 5(a) show the SNPs with and without the GO coating. The thickness of the GO layer is about 30 nm. Fig. 5(b) presents the measured reflection spectra of the device when the SNPs is bare (black), coated with a GO layer (red), and functionalized with anti-ErbB2 antibodies.

Fig. 5(c) shows the reflectance spectra recorded when the samples containing ErbB2 are injected into the nanofluidic channel. The sample solution is prepared by dissolved ErbB2 in a phosphate-buffered saline (PBS; pH = 7.4) solution at six concentrations (0.01 nM, 0.1 nM, 0.5 nM, 1 nM, 10 nM, and 100 nM). As the concentration of ErbB2 increases, the resonance dip in the reflection spectrum red shifts to longer wavelengths. The total shift of the resonance wavelength is approximately 6 nm when the ErbB2 concentration increases from 0.01 nM to 100 nM. The dose response for the detection of ErbB2 biomarker is shown in Fig. 5(d). The experiment with a specific ErbB2 concentration repeats five times and the error bars represent the standard deviation of resonance wavelength for each concentration. The sensitivity of optical sensors can be estimated using \( S_s = \frac{\Delta \lambda}{\Delta c} \), where \( \Delta c \) is the corresponding concentration difference. The SNP-based biosensor exhibits a sensitivity of 2 nm/nM for ErbB2.

Table 1 compares the key performance of this biosensor with some recently reported label-free biosensors for the detection of ErbB2. With regards to the sensitivity, the SNP-based biosensor outperforms the reported ring resonator biosensor \( (S_s = 0.12\text{ nm/nM}) \) and distributed feedback grating \( (S_s = 2 \times 10^{-3}\text{ nm/nM}) \) biosensor (Gohring et al., 2010; Retolaza et al., 2016). The limit of detection (LOD) of the SNP-based biosensor is found to be 0.7 ng mL\(^{-1}\) using the 3 \( \sigma \) criterion (expressed in concentration units), where \( \sigma \) represents the standard deviation of five-time measurements for the PBS solution. Compared to the other two optical biosensors, the SNP-based biosensor provides a lower LOD. In addition to the high refractive index sensitivity of the metasurface, the lateral flow-through design contributes significantly to the increased sensitivity and the lowered LOD by offering a useful

<table>
<thead>
<tr>
<th>Device</th>
<th>Detection type/mode</th>
<th>Sensitivity</th>
<th>Detection range (nM)</th>
<th>Limit of detection (ng mL(^{-1}))</th>
<th>Ref.</th>
</tr>
</thead>
<tbody>
<tr>
<td>Optofludic silicon metasurface</td>
<td>Guided mode resonance</td>
<td>2 nm nM(^{-1})</td>
<td>0.01 – 10</td>
<td>0.7</td>
<td>This work</td>
</tr>
<tr>
<td>Optofludic ring resonator</td>
<td>Whispering gallery mode</td>
<td>0.12 nm nM(^{-1})</td>
<td>0.14 – 3.45</td>
<td>13</td>
<td>(Gohring et al., 2010)</td>
</tr>
<tr>
<td>1D distributed feedback grating</td>
<td>1st-order diffraction mode</td>
<td>2 × 10(^{-3}) nm nM(^{-1})</td>
<td>0.028 – 138</td>
<td>14</td>
<td>(Retolaza et al., 2016)</td>
</tr>
<tr>
<td>Hydrazine-Au nanoparticle-aptamer</td>
<td>Electrochemical</td>
<td>15 × 10(^{-3}) μA nM(^{-1})</td>
<td>1 × 10(^{0}) – 1.381</td>
<td>37</td>
<td>(Zhu et al., 2013b)</td>
</tr>
<tr>
<td>Sandwich magnetomimmsensor</td>
<td>Electrochemical</td>
<td>1.4 × 10(^{3}) A nM(^{-1})</td>
<td>1.4 × 10(^{-3}) – 0.4</td>
<td>0.026</td>
<td>(Eletxigerra et al., 2015)</td>
</tr>
<tr>
<td>Capacitance based aptasensor</td>
<td>Electrochemical</td>
<td>1.7 × 10(^{-3}) pF nM(^{-1})</td>
<td>2.7 × 10(^{-3}) – 2.7 × 10(^{-2})</td>
<td>0.2</td>
<td>(Qureshi et al., 2015)</td>
</tr>
</tbody>
</table>
platform for realizing the syngeneic interaction between minute amounts of ErbB2 proteins available in the sample and the capture molecules on the surface of SNPs. Although some of the reported electrochemical biosensors exhibit lower LOD values, our device presents a wider dynamic range.

To study the selectivity of the label-free biomarker detection assay, samples containing ErbB2, ErbB3, and ErbB4 antigens were measured using the anti-ErbB2 antibody-coated metasurface. These antigens belong to the ErbB receptor tyrosine kinase family, but the ErbB3 and ErbB4 antigens are non-specific to the anti-ErbB2 antibody (Ali et al., 2016b). Fig. 5(e) shows the measured reflection spectra of seven samples with different combinations of ErbB2, ErbB3, and ErbB4 antigens (1 nM). The inset of Fig. 5(e) summarizes the resonance wavelength measured for these samples. The samples with the ErbB2 antigen and interfering molecules result in a resonance wavelength shift of 6 nm. In contrast, the samples without ErbB2 molecules show very small shift (< 1 nm) of the resonance wavelength. The results indicate that the SNP-based biosensor is selective to the target ErbB2 antigen when the SNPs are functionalized using the anti-ErbB2 antibody. As a label-free detection method, the biosensor can monitor the analyte-ligand binding process in real time. In this experiment, the reflection spectra were recorded when the ErbB2 bound to the ErbB2 antibody-coated sensor surface. Fig. 5(f) shows the temporal change of the resonance wavelength when the ErbB2 sample (0.01 nM) passes through at a flow rate of 0.3 µl/min. The simulated and experimental results for ErbB2 and anti-ErbB2 binding are compared in Fig. 5(f). The error bars show the standard deviation of resonance wavelength shift for three replicated tests. The simulation of the binding process is performed using a finite element method (FEM) model that includes the fluid dynamics of the sample inside the nanofluidic channel, the diffusion of analyte from the solution to the sensor surface, and the surface reaction process. The details of the simulation are described in the Experiment section. The binding of the ErbB2 to the surface results in an increase of λ₀ around 6 nm.

4. Conclusions

In summary, this paper demonstrates a label-free, lateral flow biosensor that combines both biomolecule detection and sample delivery functions using the SOI-based metasurface. The SNPs provide numerous nanoscale flow channels to facilitate rapid delivery of analyte to the sensor surface. The device utilizes the GMR mode of the metasurface and operates in the telecom optical wavelength band. The linewidth of the resonance is tuned by changing the angle of incidence. The biosensor exhibits the refractive index sensitivity of 720 nm/RIU. The biosensor is studied for its ability to detect the ErbB2 breast cancer biomarker. Since the SNP-based biosensor can be fabricated using the CMOS-compatible process, the device is amendable to integration with a wide variety of lab-on-a-chip components. We envision that the biosensor will enable rapid and quantitative analysis in point-of-care applications, such as disease diagnosis, drug test, and pathogen detection.

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Appendix A. Supporting information

Supplementary data associated with this article can be found in the online version at http://dx.doi.org/10.1016/j.bios.2018.02.038.