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Integrated photonic glucose biosensor using a vertically coupled microring resonator in polymers

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ABSTRACT

A photonic glucose biosensor incorporating a vertically coupled polymeric microring resonator was proposed and accomplished. The concentration of a glucose solution was estimated by observing the shift in the resonant wavelength of the resonator. For achieving higher sensitivity the contrast between the effective refractive index of the polymeric waveguide and that of the analyte was minimized. Actually, the effective refractive index of the polymeric waveguide ($n = \sim 1.390$) was substantially close to that ($n = \sim 1.333$) of the fresh solution with no glucose. The fabricated resonator sensor with the free spectral range of 0.66 nm yielded a sensitivity of ~280 pm/(g/dL), which corresponds to ~200 nm/RIU (refractive index units) as a refractometric sensor, and provided a detection limit of refractive index change on the order of 10^{-5} RIU.

High performance glucose sensors capable of monitoring the blood sugar level are indispensable to an enormous number of diabetic patients [1]. Especially an integrated photonic glucose sensor has received a lot of attention rendering many advantages including robustness and compactness, high surface specificity using surface chemical modifications, easy patterning of reagents, and high integration capability with other electronic/photonic devices. In addition, it is compatible with microfluidic handling and can accommodate a multi-channel (multi-analyte) sensing. In the meantime, compact photonic biosensors incorporating a microring resonator were preferentially researched in view of such merits as high sensitivity, small footprint, flexible integration, and affordable mass production [2–6].

In this paper, we attempted to develop an integrated photonic glucose biosensor exploiting a vertically coupled microring resonator in polymeric waveguides. The concentration of the aqueous glucose solution was estimated by observing the shift in the resonant wavelength of the resonator. In order to boost the sensitivity, a polymeric waveguide system was devised in such a way that the difference between its effective refractive index (*RI*) and the *RI* of the target analyte was minimized. A ring waveguide with a pedestal structure, where the residual cladding adjacent to the ring itself was excessively etched away, was also introduced to enlarge the contact area between the ring and the analyte for achieving higher sensitivity.

The proposed glucose biosensor resorting to a polymeric microring resonator is illustrated in Fig. 1. The configuration of a vertical bus-to-ring coupling was adopted instead of that of a lateral coupling, since the former is less affected by the existence of the target analyte compared to the latter [4]. The sensor device involves a pedestal type of ring [5], which is vertically connected to an embedded bus waveguide serving as input/output ports. Light is initially launched to the input port of the lower bus and partially coupled to the upper ring, and continues to travel along the ring. The launched light is interfered with the coupled light circulating around the ring, so that a periodic band-stop filtering characteristic is attained at the output of the bus. Here the period is denoted as a free spectral range (FSR) and the resonant wavelength is given by $\lambda_c = n_{eff} 2\pi R/m$, where n_{eff} is the effective *RI* of the ring's guided mode, R is the ring radius, and *m is an integer.*

As implied by the device structure illustrated in Fig. 1, the target analyte comprising the aqueous solution plays the role of the top cladding of the ring waveguide. Its *RI* is denoted as n_a. A change in the *RI* of the analyte Δn induced by the change in the concentration thereof leads to a variation Δn_{eff} in the effective *RI* of the ring itself. As a result, the resonant wavelength of the resonator sensor is shifted by $\Delta\lambda_c = \Delta n_{eff} 2\pi R/m$, indicating there is a linear relationship between $\Delta\lambda_c$ and Δn_{eff} and therefore the glucose concentration can be estimated from $\Delta\lambda_c$ [5,6]. To elevate the sensitivity of such a refractometric biosensor, Δn_{eff} available from the ring waveguide should be maximized for a given variation in the glucose concentration. For the case of sensors relying on a source light with a fixed wavelength the glucose concentration may be also extracted by checking on the variation in its output optical power,

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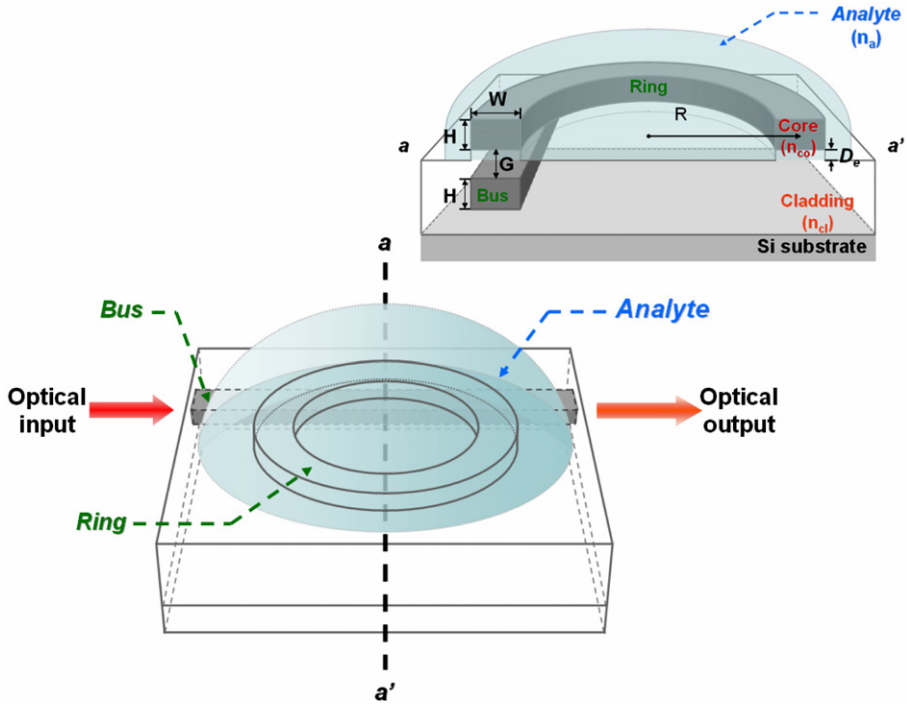

Fig. 1. Schematic configuration of the proposed glucose biosensor.

where the sensitivity could be potentially high but the dynamic range limited [5]. Meanwhile, the proposed wavelength shift method is able to offer wider dynamic range but the sensitivity may not be satisfactory [6].

The sensitivity of the proposed sensor is expected to be inversely proportional to the magnitude of the initial difference between the *RI* of the target analyte and the effective *RI* of the ring waveguide acting as sensing part [7]. For instance, a ring resonator sensor in silicon, which accompanies relatively huge index-difference between the silicon ($n = \sim 3.5$) and the target analyte ($n = \sim 1.3$), exhibited relatively low sensitivity [8]. We have first analyzed the device sensitivity for different values of the initial contrast between the effective *RI* of the ring and the *RI* of the aqueous solution with no glucose with the help of the film mode-matching method. The simulation parameters associated with the waveguide were as follows: the stripe wide $W = 2.8$ μm, the height $H = 2$ μm, and the *RI* of the top cladding $n_{a0} = 1.3334$. Here the rate of change of the *RI* of the solution with respect to the glucose concentration was assumed to be 0.0014/(g/dL) [9]. Fig. 2 shows the calculated shift in the resonant wavelength versus the glucose concentration for various initial index contrasts (Δn_d) between the ring (n_{eff0}) and the top cladding (n_{a0}), revealing that the sensitivity varied from 51 pm/(g/dL) to 276 pm/(g/dL) as Δn_d was altered from 0.14 to 0.03. Here smaller Δn_d was proved to lead to higher sensitivity, which is however meaningful only to the extent that the spectral resolution or the detection limit of the sensor is not severely degraded. Actually, the confinement of the guided mode of the ring waveguide becomes weakened with smaller index contrast. Hence, its propagation loss is elevated stemming primarily from the bending loss in addition to the absorption loss and the scattering loss, and accordingly the finesse of the ring resonator is diminished [10]. Consequently it is to be noted that there is a trade-off between the index contrast and the device performance.

For boosting the sensitivity, it is also essential to expand the sensing region determined by the contact between the ring and the analyte and to reinforce the influence of the analyte upon the effective *RI* of the ring [11]. Toward that end, a pedestal ring structure was exploited as displayed in Fig. 1, where the residual lower cladding in the vicinity of the side wall of the core of the ring waveguide was excessively etched away. In this way, the influence of the top cladding on the properties of the guided mode of the ring was strengthened [12]. Moreover, the quality factor available from the pedestal ring was heightened as a result of lower propagation loss resulting from stronger mode confinement [13].

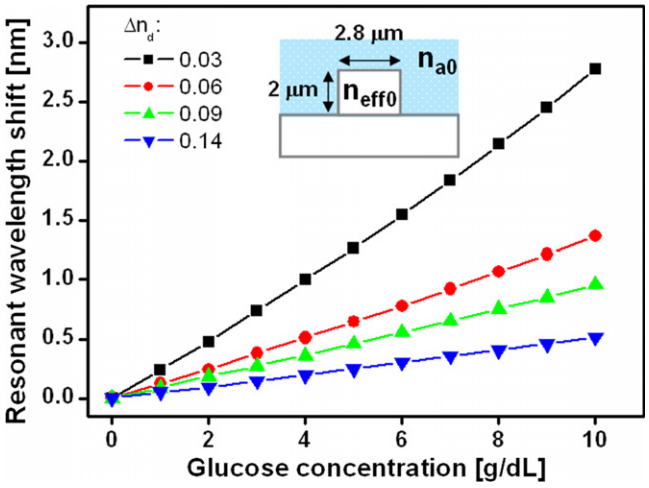

Fig. 2. Theoretical sensitivity of the resonator sensor for different initial index contrasts between the ring and the glucose solution.

The design of the resonator sensor was performed via the beam propagation method, and its results are summarized as follows: the RI of the core and the lower cladding was $n_{co} = 1.430$ and $n_{cl} = 1.375$, respectively. The stripe width and the height of the ring and bus waveguide were selected to be $W = 2.8$ μm and $H = 2$ μm respectively, thereby ensuring a single-mode operation. The depth of the over-etching of the cladding layer was $De = 0.5$ μm. And the *RI* of the top cladding of the ring was supposed to be ~1.333 for the

zero glucose concentration. The effective *RI* of the ring was calculated to be 1.390 and the corresponding initial index contrast Δn_d was as small as 0.057. Here the effective *RI* was estimated by using the OlympIOs, a simulation tool based on the beam propagation method [14]. And the vertical gap between the ring and the bus was $G = 0.9$ μm, and the ring radius was chosen to be $R = 400$ μm considering the bending loss. The change in the effective RI of the ring with the RI of the top cladding was examined by employing the Selene Interface of OlympIOs. From the analysis of the ring resonator through the transfer matrix method, the rate of change of the ring's effective *RI* Δn_{eff} was found to be ~0.22/RIU, which is equivalent to the sensor sensitivity of ~290 pm/(g/dL).

The polymeric biosensor was created by adopting the standard procedure used for planar lightwave circuits as described in Fig. 3. The lower cladding was first formed by spin-coating a LFR-S708U ($n = 1.375$) polymer from ChemOptics on a silicon substrate, and the bus waveguide pattern was produced in a photoresist layer via the lithography process. The lower cladding was then selectively dry-etched to transfer the waveguide pattern in photoresist onto it. And a ZPU13-430 polymer ($n = 1.430$) was spin coated atop the lower cladding with the residual layer removed by appropriate dry-etching, whereupon another 0.9 μm thick film of LFR-S708U polymer was subsequently formed. Here the core of the embedded bus was 2 μm high and 2.8 μm wide. Next the second core layer of 2.0 μm thickness belonging to the ring waveguide was made by spin-coating the ZPU13-430 polymer. Likewise, the ring waveguide pattern with a radius of 400 μm was created via the lithography followed by the selective dry-etching. The lower cladding was etched down below the core by the depth of 0.5 μm so that the side wall of the ring was completely exposed. Finally, the end facets were prepared by dicing. The micrograph of the completed polymeric sensor has been inserted to Fig. 3.

The fabricated sensor was characterized by using the measurement setup engaging a tunable laser, an optical power meter, a polarization beam splitter, and a polarizer. The transfer curve for the device was first observed for the TE polarization and plotted in Fig. 4 while a de-ionized (DI) water ($n = \sim 1.333$) was placed on the surface of the resonator sensor. For the spectral band near

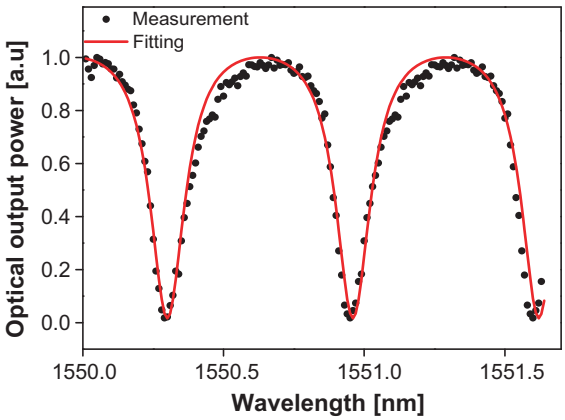

Fig. 4. Observed transfer characteristics of the proposed ring resonator.

1550 nm, the FSR was measured to be 0.66 nm, the bandwidth 0.15 nm, and the corresponding quality factor ~10,000. As shown in Fig. 4, the observed spectral response offered a good agreement with the theoretical response that was obtained by performing fitting through the transfer matrix method [15]. The measurement results for the TM polarization were nearly the same as those for the TE polarization except for a slight spectral offset. These experimental results proved that the fabricated resonator sensor functions as a periodic band-stop filter as desired.

Next the device was investigated as a glucose biosensor. At the outset, the top surface of the sensor was completely covered with a fresh DI-water solution (40 μL) containing no glucose. A discrete amount of glucose (20 μL) solution with the concentration of 500 mg/dL was successively added to the solution existing on the surface of the sensor via a micro pipette so as to alter the glucose concentration thereof from 0 to 300 mg/dL (specifically 0 mg/dL, 167 mg/dL, 250 mg/dL, 300 mg/dL), while the spectral response for different glucose concentrations was recorded with respect to that for the initial condition of zero concentration. Fig. 5a depicts the spec-

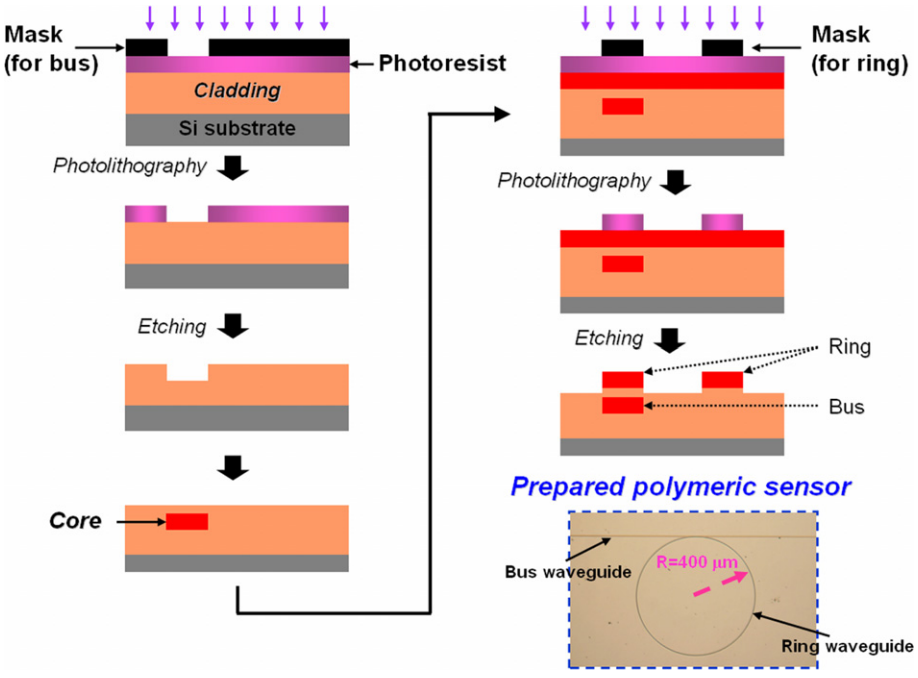

Fig. 3. Fabrication procedure for the sensor.

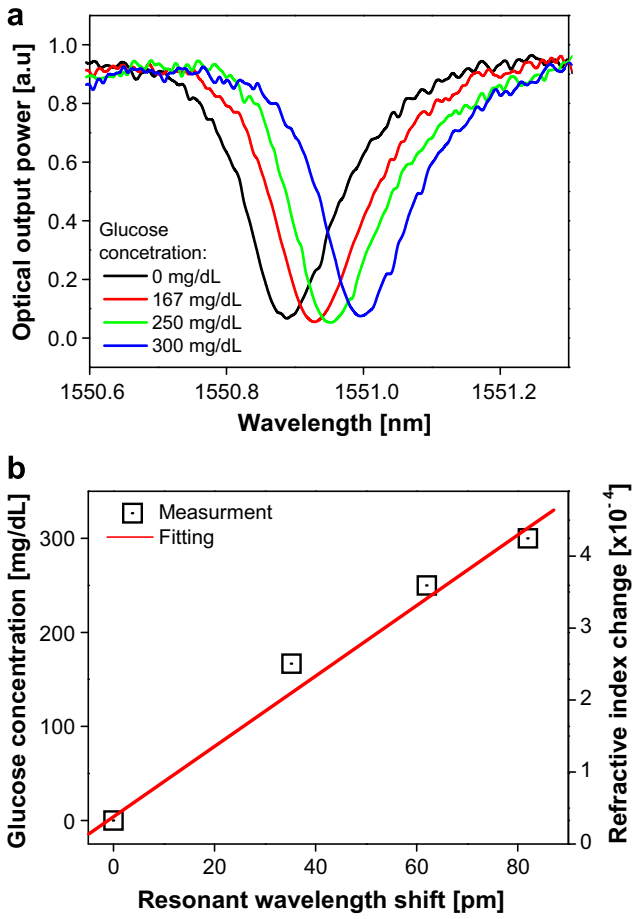

Fig. 5. Demonstrated sensor performance (a) spectral response for different glucose concentrations (b) resonant wavelength shift and *RI* change as a function of the glucose concentration.

tral response of the device in terms of the glucose concentration, indicating that the resonant wavelength was red-shifted with larger glucose concentrations. And Fig. 5b reveals the resonant wavelength shift as a function of the glucose concentration, which was linearly fitted by means of the least square method to give rise to the sensitivity of 280 pm/(g/dL). This sensitivity was in close agreement with the theoretical result (290 pm/(g/dL)) discussed earlier. The variation in the *RI* of the glucose solution as a function of the glucose concentration is displayed in Fig. 5b as well, reflecting that the proposed sensor delivered the sensitivity of as high as ~200 nm/RIU as a refractometric sensor. Our sensor is believed to demonstrate modestly enhanced sensitivity compared to other previous sensors [5,6]. Next it was evaluated in terms of the detection limit, or the resolution [3]. Assuming the spectral resolution of 1 pm available from the light source, the observed detection limit was found to be 5×10^{-6} RIU, which is comparable to that of other devices [4,7,8].

In summary, an integrated photonic glucose biosensor drawing upon a vertically coupled microring resonator in polymeric waveguides was presented. Its sensitivity was enhanced by diminishing the index contrast between the waveguides and the target analyte. It can be operated by taking advantage of the variation in the output optical power for a fixed light wavelength as well. In view of its practical application to diabetic patients, the sensor is required to provide an exclusive selectivity to the glucose since the blood is a complex, inhomogeneous mixture of various substances (proteins, lipids, amino acids, etc.) in addition to the glucose and water [16]. Such approaches as the antibody–antigen scheme will be potentially helpful for enhancing the selectivity of the sensor [2,5].

Acknowledgement

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