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Optical Biosensors Based on Multimode Interference and Microring Resonator Structures: A Personal Perspective

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Abstract: We review our recent works on optical biosensors based on microring resonators (MRR) integrated with 4x4 multimode interference (MMI) couplers for multichannel and highly sensitive chemical and biological sensors. Our proposed sensor structures have advantages of compactness, high sensitivity compared with the reported sensing structures. By using the transfer matrix method (TMM) and numerical simulations, the designs of the sensor based on silicon waveguides are optimized and demonstrated in detail. We applied our structure to detect glucose and ethanol concentrations simultaneously. A high sensitivity of 9000 nm/RIU, detection limit of 2x10-4 for glucose sensing and sensitivity of 6000nm/RIU, detection limit of 1.3x10-5 for ethanol sensing are achieved.

Keywords: Biological sensors, chemical sensors, optical microring resonators, high sensitivity, multimode interference, transfer matrix method, beam propagation method (BPM), multichannel sensor.

1. Introduction

Current approaches to the real time analysis of chemical and biological sensing applications utilize systematic approaches such as mass spectrometry for detection. Such systems are expensive, heavy and cannot monolithically integrated in one single chip [1]. Electronic sensors use metallic probes which produces electro-magnetic noise, which can disturb the electro-magnetic field being measured. This can be avoided in the case of using integrated optical sensors. Integrated optical sensors are very attractive due to their advantages of high sensitivity and ultra-wide bandwidth, low detection limit, compactness and immunity to electromagnetic interference [2, 3].

Optical sensors have been used widely in many applications such as biomedical research, healthcare and environmental monitoring. Typically, detection can be made by the optical absorption of the analytes, optic spectroscopy or the refractive index change [1]. The two former methods can be directly obtained by measuring

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optical intensity. The third method is to monitor various chemical and biological systems via sensing of the change in refractive index [4].

Optical waveguide devices can perform as refractive index sensors particularly when the analyte becomes a physical part of the device, such as waveguide cladding. In this case, the evanescent portion of the guided mode within the cladding will overlap and interact with the analyte. The measurement of the refractive index change of the guided mode of the optical waveguides requires a special structure to convert the refractive index change into detectable signals. A number of refractive index sensors based on optical waveguide structures have been reported, including Bragg grating sensors, directional coupler sensors, Mach-Zehnder interferometer (MZI) sensors, microring resonator sensors and surface plasmon resonance sensors [1, 4-7].

Recently, the use of optical microring resonators as sensors [2, 6] is becoming one of the most attractive candidates for optical sensing applications because of its ultra-compact size and easy to realize an array of sensors with a large scale integration [8-10]. When detecting target chemicals by using microring resonator sensors, one can use a certain chemical binding on the surface. There are two ways to measure the presence of the target chemicals. One is to measure the shift of the resonant wavelength and the other is to measure the optical intensity with a fixed wavelength.

In the literature, some highly sensitive resonator sensors based on polymer and silicon microring and disk resonators have been developed [11-14]. However, multichannel sensors based on silicon waveguides and MMI structures, which have ultra-small bends due to the high refractive index contrast and are compatible with the existing CMOS fabrication technologies, are not presented much. In order to achieve multichannel capability, multiplexed single microring resonators must be used. This leads to large footprint area and low sensitivity. For example, recent results on using single microring resonators for glucose and ethanol detection showed that sensitivity of 108nm/RIU [2, 15], 200nm/RIU [16] or using microfluidics with grating for ethanol sensor with a sensitivity of 50nm/RIU [17]. Silicon waveguide based sensors has attracted much attention for realizing ultra-compact and cheap optical sensors. In addition, the reported sensors can be capable of determining only one chemical or biological element.

The sensing structures based on one microring resonator or Mach Zender interferometer can only provide a small sensitivity and single anylate detection [13]. This study presents a review on our works published in recent years for optical biosensor structures to achieve a highly sensitive and multichannel sensor.

2. Two-parameter sensor based on 4x4 MMI and resonator structure

We present a structure for achieving a highly sensitive and multichannel sensor [18]. Our structure is based on only one 4x4 multimode interference (MMI) coupler assisted microring resonators [19, 20]. The proposed sensors provide very high sensitivity compared with the conventional MZI sensors. In addition, it can measure two different and independent target chemicals and biological elements simultaneously. We investigate the use of our proposed structure to glucose and ethanol sensing at the same time. The proposed sensor based on 4x4 multimode interference and microring resonator structures is shown in Fig. 1. The two MMI couplers are identical. The two 4x4 MMI couplers have the same width W_{MMI} and length L_{MMI}.

In this structure, there are two sensing windows having lengths L_{arm1}, L_{arm2}. As with the conventional MZI sensor device, segments of two MZI arms overlap with the flow channel, forming two separate sensing regions. The other two MZI arms isolated from the analyte by the micro fluidic substrate. The MMI coupler consists of a multimode optical waveguide that can support a number of modes [21]. In order to launch and extract light from the multimode region, a number of single mode access waveguides are placed at the input and output planes. If there are N input waveguides and M output waveguides, then the device is called an NxM MMI coupler.

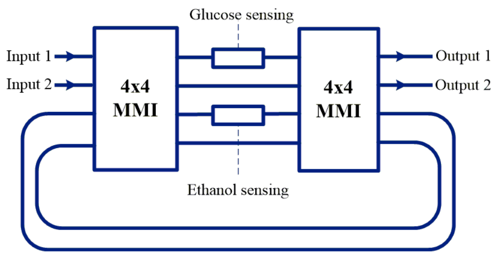


Fig. 1. Schematic of the new sensor using 4x4 MMI couplers and microring resonators.

If we choose the MMI coupler having a length of $L_{MMI} = \frac{3L_\pi}{2}$, where L_π is the beat length of the MMI coupler [22]. One can prove that the normalized optical powers transmitted through the proposed sensor at wavelengths on resonance with the microring resonators are given by [9]

$$T_1 = \left\{ \frac{\alpha_1 - \left|\cos(\frac{\Delta\varphi_1}{2})\right|}{1-\alpha_1\left|\cos(\frac{\Delta\varphi_1}{2})\right|} \right\}^2,$$

$$T_2 = \left\{ \frac{\alpha_2 - \left|\cos(\frac{\Delta\varphi_2}{2})\right|}{1-\alpha_2\left|\cos(\frac{\Delta\varphi_2}{2})\right|} \right\}^2,$$

Here $\tau_1 = \sin(\frac{\Delta\varphi_1}{2})$, $\kappa_1 = \cos(\frac{\Delta\varphi_1}{2})$ $\tau_2 = \sin(\frac{\Delta\varphi_2}{2})$, and $\kappa_2 = \cos(\frac{\Delta\varphi_2}{2})$; $\Delta\varphi_1$, $\Delta\varphi_2$ are the phase differences between two arms of the MZI, respectively; α_1, α_2 are round trip transmissions of light propagation through the two microring resonators [23].

In this study, the locations of input, output waveguides, MMI width and length are carefully designed, so the desired characteristics of the MMI coupler can be achieved. It is now shown that the proposed sensor can be realized using silicon nanowire waveguides [24, 25]. By using the numerical method, the optimal width of the MMI is calculated to be $W_{MMI} = 6\mu m$ for high performance and compact device. The core thickness is h_{co}=220nm. The access waveguide is tapered from a width of 500nm to a width of 800nm to improve device performance. It is assumed that the designs are for the transverse electric (TE) polarization at a central optical wavelength $\lambda = 1550nm$. The FDTD simulations for sensing operation when input signal is at port 1 and port 2 for glucose and ethanol sensing are shown in Fig. 2(a) and 2(b), respectively. The mask design for the whole sensor structure using CMOS technology is shown in Fig. 2(c).

The proposed structure can be viewed as a sensor with two channel sensing windows, which are separated with two power transmission characteristics T_1, T_2 and sensitivities S_1, S_2. When the analyte is presented, the resonance wavelengths are shifted. As the result, the proposed sensors are able to monitor two target chemicals simultaneously and their sensitivities can be expressed by:

$$S_1 = \frac{\partial\lambda_1}{\partial n_c}, \quad S_2 = \frac{\partial\lambda_2}{\partial n_c},$$

where λ_1 and λ_2 are resonance wavelengths of the transmissions at output 1 and 2, respectively.

For the conventional sensor based on MZI structure, the relative phase shift $\Delta\varphi$ between two MZI arms and the optical power transmitted through the MZI can be made a function of the environmental refractive index, via the modal effective index n_{eff}. The transmission at the bar port of the MZI structure can be given by [1]

$$T_{MZI} = \cos^2\left(\frac{\Delta\varphi}{2}\right) \quad (4)$$

where $\Delta\varphi = 2\pi L_{arm}(n_{eff,a} - n_{eff,0})/\lambda$, L_{arm} is the interaction length of the MZI arm, $n_{eff,a}$ is effective refractive index in the interaction arm when the ambient analyte is presented and $n_{eff,0}$ is effective refractive index of the reference arm.

The sensitivity S_{MZI} of the MZI sensor is defined as a change in normalized transmission per unit change in the refractive index and can be expressed as

$$S_{MZI} = \frac{\partial T_{MZI}}{\partial n_c}$$

where n_c is the cover medium refractive index or the refractive index of the analyte. The sensitivity of the MZI sensor can be rewritten by

$$S_{MZI} = \frac{\partial T_{MZI}}{\partial n_c} = \frac{\partial T_{MZI}}{\partial n_{eff,a}} \frac{\partial n_{eff,a}}{\partial n_c}$$

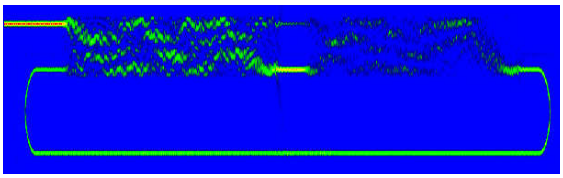
The waveguide sensitivity parameter $\frac{\partial n_{eff,a}}{\partial n_c}$ can be calculated using the variation theorem for optical waveguides [1]:

$$\frac{\partial n_{eff,a}}{\partial n_c} = \frac{\frac{n_c}{n_{eff,a}} \iint_{analyte} |E_a(x,y)|^2 dxdy}{\iint_{\infty} |E_a(x,y)|^2 dxdy}$$

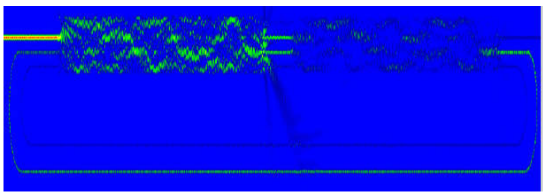
Where $E_a(x,y)$ is the transverse field profile of the optical mode within the sensing region, calculated assuming a dielectric material with index n_c occupies the appropriate part of the cross-section. The integral in the numerator is carried out over the fraction of the waveguide cross-section occupied by the analyte and the integral in the denominator is carried out over the whole cross-section.

For sensing applications, sensor should have steeper slopes on the transmission and phase shift curve for higher sensitivity. From **Error! Reference source not found.** and **Error! Reference source not found.**, we see that the sensitivity of the MZI sensor is maximized at phase shift $\Delta\varphi = 0.5\pi$. Therefore, the sensitivity of the MZI sensor can be enhanced by increasing the sensing window length L_a or increasing the waveguide sensitivity factor $\frac{\partial n_{eff,a}}{\partial n_c}$, which can be obtained by properly designing optical waveguide structure. In this chapter, we present a new sensor structure based on microring resonators for very high sensitive and multi-channel sensing applications.

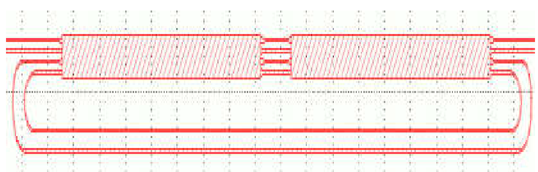
From equations **Error! Reference source not found.** and **Error! Reference source not found.**, the ratio of the sensitivities of the proposed sensor and the conventional MZI sensor can be numerically evaluated. The sensitivity enhancement factor S_1/S_{MZI} can be calculated for values of α_1 between 0 and 1 is plotted in Fig. 3. For $\alpha_1 = 0.99$, an enhancement factor of approximately 10 is obtained. The similar results can be achieved for other sensing arms.



(a) Input 1, glucose sensing



(b) Input 2, Ethanol sensing



(c) Mask design

Fig. 2. FDTD simulations for two-channel sensors (a) glucose, (b) Ethanol and (c) mask design [18].

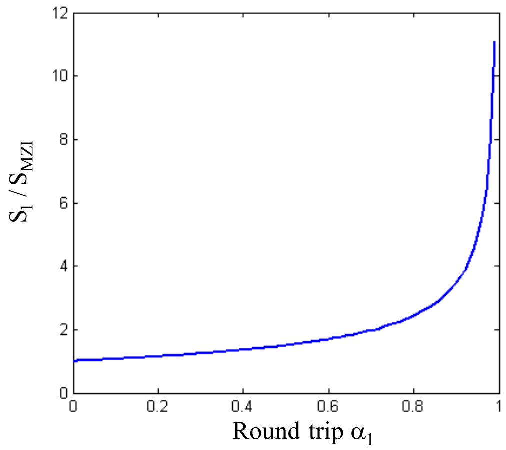


Fig. 3. Sensitivity enhancement factor for the proposed sensor, calculated with the first sensing arm.

In general, our proposed structure can be used for detection of chemical and biological elements by using both surface and homogeneous mechanisms. Without loss of generality, we applied our structure to detection of glucose and ethanol sensing as an example. The refractive indexes of the glucose ($n_{glucose}$) and ethanol (n_{EtOH}) can be calculated from the concentration (C%) based on experimental results at wavelength 1550nm by [26-28]

$$n_{glucose} = 0.2015x[C] + 1.3292 \quad (8)$$

$$n_{EtOH} = 1.3292 + a[C] + b[C]^2 \quad (9)$$

where $a = 8.4535x10^{-4}$ and $b = -4.8294x10^{-6}$.

By measuring the resonance wavelength shift ($\Delta\lambda$), the glucose concentration is detected. The sensitivity of the glucose sensor can be calculated by [18]

$$S_{glucose} = \frac{\Delta\lambda}{\Delta n} = 9000(nm/RIU) \quad (10)$$

Our sensor provides the sensitivity of 9000 nm/RIU compared with a sensitivity of 170nm/RIU [29].

In addition to the sensitivity, the detection limit (DL) is another important parameter. For the refractive index sensing, the DL presents for the smallest ambient refractive index change, which can be accurately measured. In our sensor design, we use the optical refractometer with a resolution of 20pm, the detection limit of our sensor is calculated to be $2x10^{-4}$, compared with a detection limit of $1.78x10^{-5}$ of single microring resonator sensor [30]. The sensitivity of the ethanol sensor is calculated to be $S_{EtOH} = 6000(nm/RIU)$ and detection limit is $1.3x10^{-5}$.

It is noted that silicon waveguides are highly sensitive to temperature fluctuations due to the high thermo-optic coefficient (TOC) of silicon ($TOC_{Si} = 1.86x10^{-4}K^{-1}$). As a result, the sensing performance will be affected due to the phase drift. In order to overcome the effect of the temperature and phase fluctuations, we can use some approaches including of both active and passive methods. For example, the local heating of silicon itself to dynamically compensate for any temperature fluctuations [31], material cladding with negative thermo-optic coefficient [32-35], MZI cascading intensity interrogation [14], control of the thermal drift by tailoring the degree of optical confinement in silicon waveguides with different waveguide widths [36], ultra-thin silicon waveguides [37] can be used for reducing the thermal drift.

3. Optical biosensor based on two microring resonators

A schematic of the structure is shown in Fig. 8. The proposed structure contains one 4x4 MMI coupler, where a_i, b_i (i=1,...,4) are complex amplitudes at the input and output waveguides. Two microring resonators are used in two output ports [38].

It was shown that this structure can create Fano resonance, CRIT and CRIA at the same time [19]. We can control the Fano line shape by changing the radius R1 and R2 or the coupling coefficients of the couplers used in microring resonators. Here, microring resonator with radius R1 is used for sensing region and microring with R2 for reference region. The analyte will be covered around the cladding of the optical waveguide and therefore causing the change in effective refractive index and output spectrum of the device. By measuring the shift of the resonance wavelength, we can determine and estimate the concentration of the glucose.

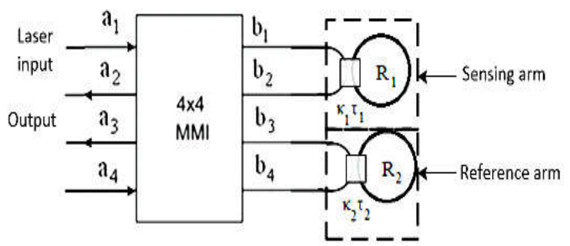


Fig. 8 Schematic diagram of a 4x4 MMI coupler based sensor

In this study, we use homogeneous sensing mechanism. where κ_1 and τ_1 are the cross coupling coefficient and transmission coupling coefficient of the coupler 1; α_1 is the loss factor of the field after one round trip through the microring resonator; $\varphi_1 = 2\pi n_{eff} L_{R1} / \lambda$ is the round trip phase, n_{eff} is the effective index and L_{R1} is the microring resonator length. The normalized transmitted power at the output waveguide is:

$$T_1 = \left|\frac{b_2}{b_1}\right|^2 = \frac{\alpha_1^2 - 2\alpha_1\tau_1\cos(\varphi_1) + \tau_1^2}{1 - 2\alpha_1\tau_1\cos(\varphi) + \alpha_1^2\tau_1^2} \quad (12)$$

When light is passed through the input port of the microring resonator, all of the light are received at the through port except for the wavelength which satisfies the resonance conditions:

$$m\lambda_r = n_{eff} L_{R1} = n_{eff}(\pi R_1) \quad (13)$$

$$m\lambda_r = n_{eff} L_{R2} = n_{eff}(\pi R_2) \quad (14)$$

where λ_r is the resonance wavelength and m is an integer representing the order of the resonance. The operation of the sensor using microring resonators is based on the shift of resonance wavelength. A small change in the effective index n_{eff} will result in a change in the resonance wavelength. The change in the effective index is due to a variation of ambient refractive index (n_a) caused by the presence of the analytes in the microring. The sensitivity of the microring resonator sensor is defined as [9, 39]

$$S = \frac{\partial\lambda_r}{\partial n_a} = \frac{\partial\lambda_r}{\partial n_{eff}}\frac{\partial n_{eff}}{\partial n_a} = \frac{\partial\lambda_r}{\partial n_{eff}}(S_W)\{nm/RIU\} \quad (15)$$

where $S_W = \frac{\partial n_{eff}}{\partial n_a}$ is the waveguide sensitivity, that depends only on the waveguide design and is a constant for a given waveguide structure. RIU is refractive index unit.

Another important figure of merit for sensing applications is the detection limit (DL) δn_a. It can be defined as

$$DL = \delta n_a \sim \frac{\lambda_r}{SQ} \sim \frac{R_{OSA}}{S}\{RIU\} \quad (16)$$

where Q is the quality factor of the microring resonator, R_{OSA} is the resolution of optical spectral analyzer [40-42]. It is desirable to have a small refractive index resolution, in which a small ambient index change can be detected. Therefore, high Q factor and sensitivity S are necessary.

We investigate the effect of ring radius on the sensing performance, the ratio of the two ring radii is defined as $a = \frac{R_2}{R_1}$, where a<1. The sensitivity of the proposed sensor is calculated by

$$S = \frac{\lambda_{shift}}{\Delta n_a} = \frac{1}{1-a}\frac{\Delta\lambda}{\Delta n_a} \quad (17)$$

$$LOD = \frac{\Delta n_{eff}}{n_{eff}} = \frac{1-a}{a}\frac{\lambda}{2\pi R_2 n_{eff}} \quad (18)$$

It is obvious that the sensitivity of the proposed structure is 1/(1-a) times than that of a sensor based on single microring resonator [43]. When a=R2/R1 approaches unity, the sensitivity of the proposed structure is much higher than that of the conventional one as shown in Fig. 9.

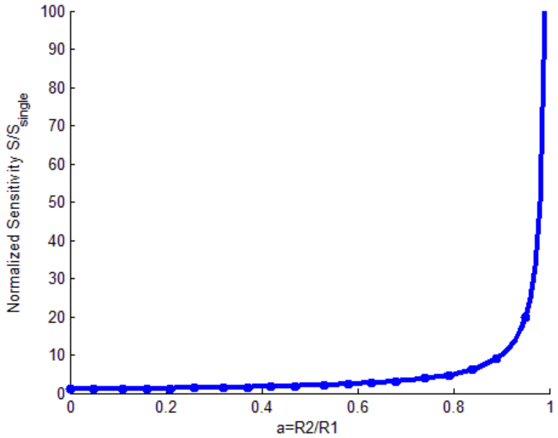


Fig. 9. Comparison of sensivity of the proposed structure with the sensitivity of the single microring sensor at different ratio between two ring radii

Now we investigate the behavior of our devices when the radius of two microring resonators is different. For example, we choose $R_1 = 20\mu m$ and $R_2 = 10\mu m$, a=0.5 and $\alpha_1 = \alpha_2 = 0.98$. It is assumed that 3dB couplers are used at the microring resonators 1 and 2. The glucose solutions with concentrations of 0%, 0.2% and 0.4% are induced to the device. For each 0.2% increment of the glucose concentration, the resonance wavelength shifts of about 800nm is achieved. This is a double higher order than that of the recent conventional sensor based on single microring resonator [27, 29].

By measuring the resonance wavelength shift ($\Delta\lambda$), the glucose concentration is detected. The sensitivity of the sensor can be calculated by [38]

$$S = \frac{\Delta\lambda}{\Delta n} = 721(\text{nm/ RIU}) \qquad (19)$$

4. Conclusions

We have presented a review on our sensor structures based on the integration of 4x4 multimode interference structure and microring resonators. The proposed sensor structures can detect two chemical or biological elements simultaneously. Our sensor structure can be realized on silicon photonics that has advantages of compatibility with CMOS fabrication technology and compactness. It has been shown that our proposed sensors can provide a very high sensitivity compared with the conventional MZI sensor.

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Cảm biến quang y sinh sử dụng cấu trúc vi cộng hưởng kết hợp với bộ ghép giao thoa đa mode

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Tóm tắt: Bài báo trình bày một số kết quả gần đây của tác giả về thiết kế một số cấu trúc cảm biến quang tích hợp y sinh mới sử dụng cấu trúc vi cộng hưởng kết hợp với cấu trúc giao thoa đa mode. Cấu trúc cảm biến sử dụng bộ ghép giao thoa đa mode 4 cổng vào, 4 cổng ra có thể đo đa kênh với độ nhạy cao, giới hạn đo thấp. Cấu trúc cảm biến đề xuất của tác giả có ưu điểm kích thước nhỏ gọn, phù hợp với chế tạo dùng công nghệ vi mạch hiện nay nên giá thành rẻ nếu chế tạo hàng loạt. Sử dụng phương pháp ma trận truyền dẫn và mô phỏng số, tác giả thiết kế tối ưu cấu trúc sử dụng ống dẫn sóng silic. Sử dụng cấu trúc đề xuất áp dụng cho phát hiện và xác định nồng độ glucose, ethanol cho thấy độ nhạy 9000nm/RIU, giới hạn đo $2x10^{-4}$ đối với cảm biến glucose và độ nhạy 6000nm/RIU, giới hạn đo $1,3x10^{-5}$ đối với ethanol có thể đạt được.

Từ khóa: Cảm biến y sinh, vi cộng hưởng quang, độ nhạy cao, đo đa kênh, cấu trúc giao thoa đa mode, phương pháp mô phỏng số.