Analysis of Integrated Optofluidic Lab-on-a-Chip Sensor Based on Refractive Index and Absorbance Sensing

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Abstract—The analysis of a fully integrated optofluidic lab-on-a-chip sensor is presented in this paper. This device is comprised of collinear input and output waveguides that are separated by a microfluidic channel. When light is passed through the analyte contained in the fluidic gap, optical power loss occurs owing to absorption of light. Apart from absorption, a mode-mismatch between the input and output waveguides occurs when the light propagates through the fluidic gap. The degree of mode-mismatch and quantum of optical power loss due to absorption of light by the fluid form the basis of our analysis. This sensor can detect changes in refractive index and changes in concentration of species contained in the analyte. The sensitivity to detect minute changes depends on many parameters. The parameters that influence the sensitivity of the sensor are mode spot size, refractive index of the fluid, molar concentration of the species contained in the analyte, width of the fluidic gap, and waveguide geometry. By correlating various parameters, an optimal fluidic gap distance corresponding to a particular mode spot size that achieves the best sensitivity is determined both for refractive index and absorbance-based sensing.

Index Terms—Absorbance, mode-mismatch, optofluidics, refractive index, sensitivity.

I. INTRODUCTION

ONE of the challenges of lab-on-a-chip system, especially for optofluidic system is the total integration of all the fluidic and optical components into a miniaturized micro-chip so that such lab-on-a-chip sensor can be used right at the point-of-care. Optofluidics is a new branch within photonics that attempts to unify concepts from optics and microfluidics [1]. Unification of photonics and microfluidics enables us to carry out analysis of fluids including human physiological fluids through highly sensitive optical sensing devices [2]–[5]. Some of the previous experimental works have focused on the integration of fluorescence and absorbance based detection schemes with fluidics. A common structural motif used in many of these works, is a fluidic gap separating two devices, one of which serves as the source of light and the other for collecting and detecting the light modulated by the fluidic gap. Previous authors have considered sensitivity analysis of integrated optical detection from the point of evanescent sensing [6]–[14]. In contrast we propose waveguide-gap structure details of which are shown in Figs. 1(a) and (b). The light when propagates through the input and output waveguide, the field profile remains unchanged and hence mode-spot size (mode-spot size is measured by taking the width of the field profile at the point where field drops to 1/e\textsuperscript{th} of its peak value) remains the same. However when the light propagates through the bulk of the analyte in the fluidic gap the mode-spot sizes enlarges. In this article we refer to mode-mismatch as the difference between mode spot sizes of the field profiles of the light entering and exiting the fluidic gap. The fluidic gap, containing the analyte, which is the molecule/material to be sensed, modifies the coupling of light from the input to the output waveguide. This is because of the mode-mismatch. Thus mode-mismatch forms an important signal transduction and can be used for analysis of such devices. The extent of the mode-mismatch depends on the device parameters such as fluidic gap distance, refractive index of the fluid, the mode spot size and so on. From the point of view of sensor design, one is interested in choosing the input and output waveguide parameters and the fluidic gap width, which maximizes the sensitivity. The values of these parameters for optimization are different for different detection techniques and are important considerations for the sensor design. In this paper we have discussed the effect of device parameters on the sensitivity of detection using the waveguide-gap structure for high throughput micro-refractometry and absorbance sensing. In refractometric or absorbance sensing, we rely on the refractive index or absorbance change caused due to variations in the molar concentration of analyte. For instance the amount tissue glucose level in blood plasma is an important indicator for detection of diabetes. The mechanism of detection of tissue glucose levels is explained in [5]. We have considered oxy-hemoglobin HbO\textsubscript{2} as an illustrative example as the analyte for the analysis described here. As pointed out earlier in this section, the considerations for performance optimizations
required for different techniques, such as changes in refractive index or absorbance, are different and one would expect the optimal sensor design for one technique to be different from that for another technique. Indeed, as we show in this paper, the optimum device parameters are different depending on whether refractometry or absorbance is the choice of measurement. In this context it may be said, that, previous authors have fabricated various devices based on refractive index [6]–[9] and absorbance based sensing [10]–[12]. In earlier works analysis of integrated optical detection from the point of evanescent sensing [13], [14] have been carried out. In this work we have focused on design and analysis of refractive-index and absorbance-based sensing, where trajectory of light is guided through the waveguides and passed through bulk of analyte. This work will prove to be useful for fabricating such sensors with optimized sensitivity.

Our proposed lab-on-a-chip device, consists of collinear input and output integrated optical waveguides separated by an intervening fluidic gap and can be built over a common glass substrate. An organic light emitting diode (OLED), if fabricated over the same glass substrate, can become an integral part of this device and can function as an internal source of light. Such an arrangement is shown in Figs. 1(a) and (b) which depict the basic concept of the miniaturized fully integrated sensor. The photo-detector (PD) which is shown in Fig. 1(b) is not shown in Fig. 1(a) as a part of the common glass substrate for the sake of clarity. Fluidic flow and transmittance of light are in mutually perpendicular directions.

Although the light source, detector and the coupling of the light from the source and to the detector are out of the scope of the current paper, we do give some qualitative comments on these issues. We have presented potentially feasible scheme wherein all the elements shown in Figs. 1(a) and (b) can be fabricated directly onto a common glass substrate. Fabrication of an organic photo-detector (OPD) and its integration with waveguide can also be achieved in the same manner as explained in [15]. The multilayer composition of OPD will consist of different organic materials as described in [16]–[18]. The OLED and the OPD have been fabricated and integrated over a common substrate [16]–[18]. The criterions for choosing detector configuration are given below.

1) The OPD should have specific spectral response and limit of detectability should be low.
2) The pixel size of the OPD should be compatible with the waveguide dimension.
3) The OPD should be amenable for monolithic integration over a common glass substrate. Also it should offer ease of fabrication, portability besides being of low cost.

The main emphasis in our work is on the following two design aspects.

1) We have designed waveguides and fluidic gap for optimization of sensitivity. This is a promising and efficient method of transmission and collection of light output from the OLED. Particularly because the out-coupling efficiency of the OLED is vastly improved due to low refractive index contrast between the OLED and waveguide [19], [20].
2) We have used mode-mismatch and absorption of optical power by the analyte as an important tool for the sensitivity analysis of the sensor.

In the following sections we detail the analysis method to determine the optimal parameters of this sensor device for refractometry and absorbance based sensing.

This paper is presented in the following sequence. In section II we have discussed the waveguide structure and evaluated the output spectrum of the waveguide. In section III the propagation of light through fluidic gap have been discussed and the optical power coupling to the output waveguide is also determined. In section IV the effect of various parameters on transmittance of optical power is described. In section V and VI the effect of various parameters on sensitivity of sensor based on two different detection techniques are presented.

II. INTEGRATED WAVEGUIDE DESIGN

Generally optically based sensors will have sources of light emission (or excitation) and a detection device to monitor the changes (as modulated by the fluid under testing) in optical properties. If these two devices are interlinked, such that, free space optics is avoided, the coupling efficiency of the light and transmission of the light can be vastly improved. This is because the refractive index contrast is low when these devices are interlinked througha waveguide structure [15], [19], [20]. It is for this reason we have proposed use of channel waveguide and its design is discussed below.

The structure of channel waveguide is shown in Fig. 2. We considered a four layer waveguide structure for the design and analysis. In the four-layer structures, the uppermost layer (first layer) is a thick glass covering, having refractive-index \( n_1 \). This layer also serves as a protection layer against environmental chemical agents. The second layer below the top protective layer is the guiding layer and has the highest refractive index \( n_2 \). The third layer is the cladding layer (or the buffer layer) having refractive index \( n_3 \), the main role of this layer is to prevent leakage of light [21] from the guiding
layer into the fourth layer, which is the glass substrate, having refractive index \( n_4 \). Four-layer structures such as this can be used to model the low-loss optical waveguides [22]–[26]. The schematic diagram of the four-layer optical integrated waveguide with the refractive indices of the various layers is shown in Fig 2.

For widespread usage of biosensing applications, it is important to have low material and manufacturing costs leading one to explore polymer based devices rather than platforms such as Silicon on oxide (SOI). A tentative process flow to fabricate such devices on glass slides may consist of constructing the cladding (buffer) layer using diffusion of ions such as Ag\(^{+}\), which produces refractive index changes of the order of 1% with respect to that of the glass substrate [27], [28]. After silver ion diffusion, the guiding layer, composed of commercially available high refractive index acrylate polymers, can be lithographically patterned to yield the waveguides, followed by encapsulation with a protective polymer/glass layer to yield the final device.

To design an integrated channel waveguide the net effective refractive index \( (N_{\text{final}}) \) must be determined. In order to solve the dispersion equation for the waveguide structure, we use the effective index method (EIM) where-by the two-dimensional channel waveguide problem is converted into a one-dimensional problem. The effective index method treats the channel waveguide as the superposition of two planar waveguides. The complete details of this waveguide design are given in [29].

The spectrum of the OLED consists of various wavelengths [15]. The propagation constant \( \beta \) for every wavelength of the OLED spectrum is determined. The intensity of the OLED spectrum at the end of the input waveguide can be determined by using (1),

\[
a_{\text{out}}(\lambda) = a_{\text{in}}(\lambda) e^{-i\beta(\lambda)L}
\]

where, \( a_{\text{out}}(\lambda) \) is the output intensity of OLED spectrum after transmittance through the input waveguide for various wavelengths \( \lambda \).

\( a_{\text{in}}(\lambda) \) is the input intensity of OLED spectrum into the input waveguide for various wavelengths \( \lambda \).

\( \beta(\lambda) \) is the final propagation constant for the wavelength.

\( L \) is the length of the input waveguide which we have considered as 1000 \( \mu \text{m} \).

The thick curve in Fig. 3 is the actual OLED spectrum measured by the spectrum analyzer [15]. This spectrum is the input spectrum which is launched into the input waveguide. The curve in dotted line in Fig. 3 represents the output spectrum from the input waveguide just before entering the fluidic gap. The curve in the dotted line is obtained from (1).

It can be seen from Fig. 3, that the two curves (in dotted lines and thick lines) are congruent. This implies that the design of waveguide and the actual output spectrum from the OLED are compatible. In other words the waveguide design supports the OLED spectrum.

### III. Determination of Optical Power Coupling Efficiency

As long as the light propagates through the waveguide, the field profile and mode spot size remains constant. However when light is allowed through free space (fluidic gap), the field profile and mode spot sizes will go on changing till it reaches output waveguide. Moreover optical properties are modified by the fluid present in the fluidic gap. These facts influence the power coupling to the output waveguide. A method to evaluate the power coupling to the output waveguide is presented in this section.

Referring to Fig. 4, the light after being guided by the input waveguide will traverse the fluidic gap. The light beam will spread while propagating through the fluidic gap and the optical mode spot size will increase. Thus there is a mode mismatch between the light entering the gap and the light exiting the fluidic gap. The extent of the mode mismatch determines the optical power coupling to the output waveguide.
The mode mismatch is dependent on the input and output waveguide geometries, the refractive index of the fluid present in the gap, and the fluidic gap distance. By evaluating this mode-mismatch, we can determine the amount of light coupling to the output waveguide. Thereby, we can find the light coupling efficiency. Once coupling efficiency is known, we can determine the sensitivity of the sensor.

Using the effective index method, we have solved the dispersion equations for the four-layer channel waveguide and the final solution reduces to a symmetric one, having effective refractive index $N_{\text{eff1}}$ for the four layers, and effective refractive index $N_{\text{eff2}}$ for the three layers on either side of four layer. Because of this symmetry, the field profile $\phi_1(x)$ of the input waveguide would be also symmetric about the propagation axis with a functional dependence on $\cos(Kx)$ in the region, $0 \leq x \leq (w/2)$ and would decay exponentially in the region, $x \geq (w/2)$ where $w$ is the width of the waveguide. This is shown in Fig. 5.

The field profile $\phi_1(x)$ of the input waveguide can be written as

$$\phi_1(x) = \begin{cases} B \cos Kx & \text{for } 0 \leq x \leq w/2 \\ C e^{-\alpha x} & \text{for } x \geq w/2 \end{cases}$$  

(2)

where, $K = \sqrt{k_0^2 N_{\text{eff1}}^2 - \beta^2}$, $\alpha = \sqrt{\beta^2 - k_0^2 N_{\text{eff2}}^2}$, $B$ and $C$ are arbitrary constants, $k_0 = 2\pi/\lambda$, $\lambda$ is the wavelength, $\beta$ is the propagation constant of the input channel waveguide, and $w$ is the width of the waveguide in $\mu$m.

$N_{\text{eff1}}$ and $N_{\text{eff2}}$ are the effective indices shown in Fig. 2.

The propagation constant $\beta$ shows functional dependence [24] on following parameters

$$\beta = f_{\beta} \{w, d, \lambda, m, n_1, n_2, n_3, n_4\}$$  

(3)

where, $w$ is the width of waveguide, $d$ is the depth of waveguide, $m$ is the order of the mode propagating through the waveguide, $n_1, n_2, n_3$ and $n_4$ are refractive indices of various layers as shown in Fig. 2.

The mode spot size $S_1$, of the input waveguide just before entering the fluidic gap is given as

$$S_1 = w + 2/\alpha.$$  

(4)

Fig. 5. Schematic diagram showing the field profile in various regions of the waveguide.

Since $\alpha = \sqrt{\beta^2 - k_0^2 N_{\text{eff2}}^2}$ and $\beta$ are functionally dependent on several parameters as mentioned in (3), and hence the functional dependence of $S_1$ can be expressed as under

$$S_1 = f_{S_1} \{w, d, \lambda, m, n_1, n_2, n_3, n_4\}$$  

(5)

When the light enters the fluidic gap, the mode spot size $S_{\text{gap}}$ of the field profile $\phi_1(x)$ will become functionally dependent on two more parameters namely, refractive index $n_f$, of the fluid, and fluidic gap distance $d_{\text{gap}}$ in addition to the parameters mentioned in (5). The functional dependence of $S_{\text{gap}}$ can be summarized as

$$S_{\text{gap}} = f_{S_{\text{gap}}} \{w, d, \lambda, n_f, m, n_1, n_2, n_3, n_4, d_{\text{gap}}\}$$  

(6)

where, $S_{\text{gap}}$ is the mode spot size of field profile $\phi_1(x)$ at various fluidic gap distances $d_{\text{gap}}$ measured from the end point of the input waveguide.

From (5) and (6), we can say that the mode spot size is functionally dependent on the propagation constant $\beta$ only as long as the light is guided within the waveguide. The moment the light enters the fluidic gap, the mode spot size $S_{\text{gap}}$ enlarges and the enlargement of the mode spot size $S_{\text{gap}}$ will depend on fluidic gap distance $d_{\text{gap}}$ travelled by the light and the refractive index $n_f$ of the fluid present in the fluidic gap. The mode spot size $S_{\text{gap}}$ will be largest (after traversing the entire fluidic gap distance) and equal to the mode spot size $S_2$ at the point where the light enters the output waveguide. Mode-mismatch ($M_s$) is defined as the difference between mode spot sizes $S_{\text{gap}}$ and $S_1$ and is written as

$$M_s = S_{\text{gap}} - S_1.$$  

(7)

The extent of mode-mismatch ($M_s$) depends on $d_{\text{gap}}$ and $n_f$ and the value of $M_s$ will be highest ($S_2 - S_1$) at the entrance of output waveguide. For any given $d_{\text{gap}}$, mode-mismatch will be influenced by $n_f$. Therefore any change in analyte either by way of changes in its content or its concentration will affect the extent of mode-mismatch. It is this dependence on $n_f$ which enables detection of changes in the composition of the analyte.

Fig. 6 shows the plot of field amplitude as a function of the mode spot size $S_{\text{gap}}$ after traversing fluidic gap distances ($d_{\text{gap}}$) of 100, 200, 300, and 400 $\mu$m, within the fluidic gap.
respectively. From each of these curves mode spot size $S_{\text{gap}}$ corresponding to respective fluidic gap distance $d_{\text{gap}}$ can be evaluated (by taking the width of each curve at the point where field drops to 1/e$^{\text{th}}$ of its peak value). In may be mentioned here that mode spot size represents the expanse of field profile of propagating light. For this plot the wavelength of light ($\lambda$) and refractive index ($n_f$) of the fluid present in the gap are kept constant at 520 nm and 1.337 respectively. The initial value of the mode spot size $S_1$ is 5 $\mu$m.

The output waveguide structure has the same geometry as that of the input waveguide. Therefore, along similar lines, we can define the field of the output waveguide $\phi_2(x)$ also as

$$\phi_2(x) = \begin{cases} D \cos K_2 x & \text{for } 0 \leq x \leq u/2 \\ E e^{-\alpha_2 x} & \text{for } x \geq u/2 \end{cases}$$  \hspace{1cm} (8)$$

where, $K_2 = \sqrt{k_0^2 n_f^2 - \beta_{\text{out}}^2}$, $\alpha_2 = \sqrt{\beta_{\text{out}}^2 - k_0^2 n_f^2}$, $D$ and $E$ are arbitrary constants, $\beta_{\text{out}}$ is the propagation constant of the output waveguide.

$\beta_{\text{out}}$ has similar functional dependence on parameters to the function mentioned in (3). However these parameters would be that of the output waveguide.

The effective coupling of field profiles $\phi_1(x)$ and $\phi_2(x)$ (with intervening fluidic gap) will determine the optical power coupling the output waveguide. The field profile $\phi_1(x)$ is determined for various values of $d_{\text{gap}}$. This is done by using beam propagation method [30], [31] based on discrete Fourier Transform (DFT). The field profile of propagating light in the gap and the mode spot size $S_{\text{gap}}$ will go on changing as per the values of $d_{\text{gap}}$. Finally the field profile at the point of coupling of the light to the output waveguide would become $\phi_2(x)$.

The coupling between the field $\phi_1(x)$ emanating from the input waveguide and the field $\phi_2(x)$ of the output waveguide is represented by the overlap integral $\tau$ and is given by the relation

$$\tau = \int \phi_1(x)\phi_2^*(x)dx.$$  \hspace{1cm} (9)$$

We then normalize (9). The normalized overlap integral $\tau_{\text{nor}}$, when evaluated is revealed to be equal to the normalized power coupling to the output waveguide, $P_{\text{nor}}$, and is given by

$$\tau_{\text{nor}} = \frac{\int_{-\infty}^{\infty} \phi_1(x)\phi_2^*(x)dx}{\sqrt{\int_{-\infty}^{\infty} \phi_1^2(x)dx} \sqrt{\int_{-\infty}^{\infty} \phi_2^2(x)dx}} = P_{\text{nor}}.$$  \hspace{1cm} (10)$$

Now using (2), (8), (9), and (10), a custom-made program has been developed to find the optical power coupling efficiency for various values of mode-mismatch. Thus we correlate the various mode spot sizes $S_{\text{gap}}$ with the distance (the fluidic gap distance, $d_{\text{gap}}$) traversed by the light corresponding to various wavelengths and refractive indices of the fluid.

The various steps involved in determining the optical power coupling to the output waveguide may be summarized as:

1) Determination of the field profiles $\phi_1(x)$ and $\phi_2(x)$.
2) Determination of the mode spot size $S_1$ of the field emanating from the input waveguide
3) Determination of the mode spot size $S_{\text{gap}}$ at various fluidic gap distances (shown in Fig. 6).

4) After obtaining the value of $S_{\text{gap}}$, the mode-mismatch $M_s$ is determined by using (7).
5) Evaluation of the normalized overlap integral $\tau_{\text{nor}}$ corresponding to various mode-mismatches.

IV. OPTICAL POWER TRANSMITTANCE

The normalized overlap integral is indicative of light coupling efficiency. Thus the amount of power coupling to the output waveguide is found. The percentage of optical power coupling to the output waveguide will depend on the following parameters.

1) Fluidic gap distance $d_{\text{gap}}$ (which separates the input and output waveguides).
2) Mode-mismatch.
3) Wavelength of the light $\lambda$.
4) Refractive index of the fluid $n_f$.
5) Absorption of light by the fluid.

In this section we describe the influence of above mentioned factors on the optical power loss which is function of fluidic gap distance. Optical power variations corresponding to various gap distances are best described through transmittance curves. Plots of these transmittance curves are shown in Figs. 7, 8, and 9. These transmittance curves show the optical power drop as the light propagates through the fluidic gap.

Fig. 7 shows the variations of optical power (normalized power) coupling to the output waveguide, as a function of fluidic gap distance $d_{\text{gap}}$, for various values of wavelengths (all the wavelengths of the OLED spectrum). For this plot the mode spot size $S_1$ is kept constant at 5 $\mu$m and the refractive index of the fluid contained in the gap is kept constant at 1.337 (the refractive index of hemoglobin at 524 nm). From Fig. 7, it can be seen that as the wavelength increases, it takes a longer time for the optical power to decay. In other words, light having longer wavelengths travels a greater distance before the optical power decays by a certain percentage, and light of shorter wavelengths decays much faster in the fluidic gap.

Fig. 8 shows the transmittance curve for mode spot sizes of varying values of $S_1$ from 5 to 23 $\mu$m. It may be mentioned
here that a mode spot size $S_1$ of 5 $\mu$m implies that, the design considerations pertaining to the waveguide are such that, it supports light propagation with mode spot size $S_1$ of 5 $\mu$m.

If mode spot size $S_1$ is to be changed, the design parameters of waveguide will have to be changed suitably.

We can clearly see that the smaller the mode spot size $S_1$, the faster the optical power decays, and the larger the mode spot size $S_1$, slower the power decays. In other words, if we have a smaller mode spot size, the fluid gap distance ($d_{gap}$) has to be much smaller. On the other hand if we have a larger mode spot size, the fluidic gap distance needs to be larger.

In section V and VI it will be shown that the mode spot size $S_1$ and fluidic gap distance $d_{gap}$ have important role to play in optimizing sensitivity of the sensor. Therefore dimensions of the waveguide and fluidic gap distances have profound effect on the sensitivity and are important design parameters. Based on the optical property to be monitored these parameters are varied to optimize the sensitivity.

**A. Effect of Absorption on the Optical Power Transmission**

In this section, we consider absorption of light by inter-vening fluid. The Beer-Lambert law is most widely used to calculate the amount of optical power absorbed for a given concentration. The Beer-Lambert law is given as

$$\varepsilon(\lambda)l = \ln\frac{P_0}{P_T}$$

where, $P_0$ is the excitation power (or the power transmitted without absorption), $P_T$ is the transmitted power after absorption, $\varepsilon(\lambda)$ is the wavelength dependent specific molar absorptivity or molar extinction coefficient in $\text{cm}^{-1}\text{M}^{-1}\text{liter}$, $c$ is the molar concentration in moles/liter, and $l$ is the optical path length traversed by the light through the fluidic gap ($d_{gap}$).

The transmitted power $P_T$ after optical absorption is given by [32]

$$P_T = P_0e^{-\varepsilon(\lambda)lc}.$$  \hspace{1cm} (12)

The power absorbed by the fluid $P_A$ is given by

$$P_A = P_0 - P_T = P_0(1 - e^{-\varepsilon(\lambda)lc}).$$  \hspace{1cm} (13)

From (13) we can infer that the optical power absorbed by the analyte will depend upon its concentration, nature of its contents, fluidic gap distance traversed by the light and wavelength of the light which is used for obtaining optical response.

**B. Case Study of Oxy-Hemoglobin**

We consider a specific case of absorption of light by hemoglobin. A number of articles in the literature are available to explain the absorption of light by hemoglobin in bulk [33]–[41]. Our sensor is a specific case involving integrated optics where the mode-mismatch plays an important role in coupling of power to the output waveguide. Therefore there is a need to incorporate optical power loss due to mode-mismatch as well as optical power loss due to absorption by the analyte (oxy-hemoglobin) in our analysis. Thus, the optical power coupling to the output waveguide will be influenced by the following two factors:

1) Mode-mismatch between the input and output waveguides

2) Power loss occurring in the fluidic gap due to absorption of light. (Absorption of light is dependent on the molar extinction coefficient and molar concentration.)

In the absence of absorption, the normalized optical power coupling to the output waveguide $P_{nor}$ is determined by using (10). When absorption is also taken into account, the result obtained through (10) gets multiplied by a factor of $\varepsilon(\lambda)l$ (from the Beer-Lambert law). Thus, the net power coupling to the output waveguide can be evaluated using (14)

$$P_{co} = P_{nor}e^{-\varepsilon(\lambda)l}c.$$  \hspace{1cm} (14)

where, $P_{co}$ is the normalized power coupling to the output waveguide taking into account both mode-mismatch and absorption, $P_{nor}$ is the normalized power coupling to the output waveguide taking into account mode-mismatch only (as evaluated from(10)) and $l$ is the optical path length traversed by the light (in our case optical path length $l$ is the fluidic gap distance $d_{gap}$).

The inferences that can be drawn from equation (7) and (13) are given below:

1) The amount of optical power coupling to the output waveguide depend upon the refractive index $n_f$ and absorption of light by the analyte.

2) The changes in the power coupling to the output waveguide forms an important optical signal transduction, and is the basis for correlating with the corresponding changes in the composition of the analyte.

The optical properties get modulated when the light is passed through an analyte. The inherent propensity of an analyte to modulate optical properties is distinctive and depends upon its composition and molar concentration. It is possible to ascertain the changes in composition or molar concentration based on the specific optical property (such as refractive index, absorbance etc.) which gets modulated in more pronounced manner. Therefore it is possible to use optical response as diagnostic tool to ascertain abnormality of an analyte by employing appropriate detection technique. For instance average normal level of hemoglobin in blood is
around 15 g/dl. Below normal levels of hemoglobin concentration may be indicative of anemia, bleeding, destruction of red blood cell, leukemia, malnutrition, and nutritional deficiencies in iron, folate, vitamin B12, or vitamin B6, whereas above normal levels of hemoglobin may be indicative of congenital heart disease, cor pulmonale, dehydration, erythrocytosis, low blood oxygen levels (hypoxia), pulmonary fibrosis, and polycythemia vera [42], [43]. The wavelength of light which is used to obtain optical response also plays important role in obtaining sensitive optical response. For instance absorption in green wavelength region is very high. In contrast absorption in infrared region is low. The changes in refractive index are more readily detectible in the infrared region [5].

Fig. 9 shows the transmittance curve of light propagating in the fluidic gap with and without absorption. The effects on the absorption due to variations in concentrations of oxyhemoglobin (i.e., 7 g/dl, 11 g/dl, and 15 g/dl) have also been brought out. The mode spot size $S_1$ is kept at 5 $\mu$m for this plot.

From Fig. 9, we can see that the higher the hemoglobin concentration in the blood, the faster the optical power drops across the fluidic gap.

V. SENSITIVITY ANALYSIS BASED ON REFRACTIVE INDEX DETECTION

In order to determine the sensitivity of the sensor, we take the slope of the transmittance curve (which is the ratio of the differential change in power coupling to the output waveguide to the differential change in refractive index of fluid contained in the gap). The power coupling to the output waveguide depends upon $\lambda$, $n_f$, $d_{gap}$, and $c$. For the refractive index based sensing, we define the sensitivity $\eta_{ref}$ of the sensor as:

$$\eta_{ref} = \frac{\partial P_{co}}{\partial n_f}$$  \hspace{1cm} (15)

where, $P_{co}$ is the normalized optical power coupling to the output waveguide, and $n_f$ is the refractive index of the fluid in the gap distance.

Fig. 10(a) shows the variation of sensitivity as a function of mode-mismatch and Fig. 10(b) depicts the variation of sensitivity as a function of fluidic gap distance for various hemoglobin concentrations.

<table>
<thead>
<tr>
<th>No.</th>
<th>Hemoglobin Concentration (g/dl)</th>
<th>Maximum Sensitivity $\eta_{ref}$ (Arb. Units)</th>
<th>Fluidic Gap Distances $d_{gap}$ ($\mu$m)</th>
<th>Refractive Index Detection Limit (RIU)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1.</td>
<td>0</td>
<td>0.00407</td>
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<td>1.22850 $\times 10^{-6}$</td>
</tr>
<tr>
<td>2.</td>
<td>7</td>
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<tr>
<td>3.</td>
<td>11</td>
<td>0.00511</td>
<td>36</td>
<td>9.78473 $\times 10^{-7}$</td>
</tr>
<tr>
<td>4.</td>
<td>15</td>
<td>0.00569</td>
<td>18</td>
<td>8.78734 $\times 10^{-7}$</td>
</tr>
</tbody>
</table>

The mode spot size $S_1$ is kept at 5 $\mu$m for the two plots shown in Figs. 10(a) and (b). We can see from Fig. 10(a) and (b) that as the hemoglobin concentration increases, the point of maximum sensitivity shifts to the left. The shift of the point of maximum sensitivity to the left is because of absorption. Thus, because of absorption, the maximum sensitivity is now achieved at fluidic gap distances less than the corresponding distances when there is no absorption. Not only is the maximum sensitivity achieved at smaller gap distances, but the degree of sensitivity is also increased as the hemoglobin concentration increases. As the concentration increases, the optical power absorption also increases. This leads to a faster change of the power coupling to the output waveguide.
Table I shows the maximum sensitivity, optimum fluidic gap distance, and refractive index detection limit measured in refractive index unit (RIU) corresponding to various hemoglobin concentrations (RIU is an unit of measuring refractive index \[7\]). The mode spot size \(S_1\) was kept at 5 \(\mu\)m for this calculation. The detection limit (RIU) is calculated from (15) by assuming, that, the PD is capable of detecting at least 5 nW change in power output.

The detectability of a PD is determined by its inherent responsivity (about 1% photon converted into photo current) \[17\]. The variations of sensitivity as function of various parameters were determined and are depicted (three dimensional plots) in Figs. 11 and 12.

Figs. 11(a) and (b) show sensitivity as a function of mode-mismatch for various mode spot sizes \(S_1\) for two hemoglobin concentrations (namely, 7 g/dl and 11 g/dl). The curve with higher sensitivity corresponds to concentration of 11 g/dl. It can be observed from Figs. 11(a) and (b) that as the mode spot size increases from 5 \(\mu\)m to 10 \(\mu\)m, the sensitivity drops. In other words, the smaller the mode spot size \(S_1\), the higher the sensitivity is. However the mode spot size \(S_1\) cannot be reduced below 5 \(\mu\)m. This is because it is very difficult to fabricate OLEDs of pixel sizes less than 5 \(\mu\)m.

Fig. 12 shows sensitivity as a function of fluidic gap distance for various wavelengths for a hemoglobin concentration of 7 g/dl when the mode spot size \(S_1\) of the field emerging from the input waveguide is kept constant at 5 \(\mu\)m. This figure shows that the highest sensitivity is achieved when the wavelength of light is 540 nm (because the molar extinction coefficient is very high at 540 nm \[44\]). Since very high sensitivity is achieved in the green part of the spectrum, we have selected a green organic light emitting diode as the light source for our analysis. The spectrum of light emitted from the fabricated OLED is shown in Fig. 3.

In Fig. 12, the curve with a higher sensitivity refers to hemoglobin concentration of 7 g/dl. The curve with lower sensitivity pertains to an imaginary case, where we assume no absorption. The imaginary case of nil absorption is shown in this plot with a view to highlight the effect of wavelength on the sensitivity of the sensor. In the absence of any absorption, there is hardly any change in the sensitivity for various wavelengths. However, absorption due to the presence of hemoglobin is highly wavelength dependent. (refer Fig. 12).

**A. Sensor Design for Refractive Index-Based Sensing**

A careful examination of Fig. 11(a) and (b) will reveal the following facts. When refractive index is the basis for detection, the sensitivity decreases with increase in mode spot size \(S_1\). Also the value of RIU increases as the mode spot size \(S_1\) is increased. In other words, as the mode spot size \(S_1\) is increased, not only does the sensitivity of the device decreases but also its ability to detect changes in refractive index also declines. If oxy-hemoglobin HbO2 is the analyte under consideration for analysis, the optimum sensitivity is obtained when \(S_1\) is 5 \(\mu\)m and \(d_{gap}\) is 40 \(\mu\)m. Even though a mode spot size less than 5 \(\mu\)m will give much improved sensitivity and detection limit, the coupling of light from the OLED to the input waveguide will become difficult. Therefore, a mode spot size \(S_1\) of 5 \(\mu\)m was chosen. If however the analyte under consideration changes, the fluidic gap distance will also have to be changed (because of differences in the molar extinction coefficients for different species) to achieve maximum sensitivity.

**VI. SENSITIVITY ANALYSIS BASED ON ABSORBANCE**

In the previous section, the sensitivity of the device was based on detection of changes in refractive index. Although in our calculation in Section V, absorption was taken into account but yet, the sensitivity was analyzed based on changes in the refractive index. Detection of abnormalities of physiological
fluids can also be based on absorbance [10]–[12]. Change in the concentration of the species in the analyte can be detected using absorbance-based sensing. For this type of detection, we define the sensitivity $\eta_{ab}$ as

$$\eta_{ab} = \frac{\partial P_{co}}{\partial c} \quad (16)$$

where, $P_{co}$ is the normalized optical power coupled to the output waveguide, after undergoing absorption by the analyte, and $c$ is the concentration of the species in g/dl.

It may be stated here that the absorption of optical power is an inevitable process in any integrated optofluidic system. Optical power absorption occurs in both types of sensing (absorbance-based and refractive-index-based). The power coupling to the output waveguide can give information about changes in refractive index and changes in absorbance.

Thus, the same sensor device can be employed to either detect changes in concentration (based on absorbance) or changes in refractive index. The distinction comes when we define the sensitivity (see (15) and (16)) and the optical property which is monitored by the detecting device. It may be mentioned here that the photocurrent response acquired from PD and its further processing by appropriate mechanism (such as neural network) [45] can be either for reading refractive index change or for absorbance change. Depending upon whether the sensor is used based on absorbance or on changes in refractive index, certain design aspects pertaining to mode spot size $S_1$ and fluidic gap distance $d_{\text{gap}}$ will have to be changed accordingly, so that, the device can operate with optimized sensitivity.

The dotted lines in Fig. 13, show the transmittance curves when the mode spot size $S_1$ is kept at $5 \, \mu m$. The thick lines represent the transmittance curve when the mode spot size $S_1$ is kept at $10 \, \mu m$. The optical path length traversed by light in the fluidic channel $d_{\text{gap}}$ is longer, when the mode spot size $S_1$ is kept at $10 \, \mu m$ (thick lines). Since the path length (fluidic gap distance $d_{\text{gap}}$) is longer, the amount of optical power absorbed by the fluid (depending upon the species concentration) will be also higher. In other words, for a given concentration of hemoglobin, the sensitivity $\eta_{ab}$ will be higher if the mode spot size $S_1$ is increased as evident from Fig. 14.

![Fig. 13. Transmittance curves for various hemoglobin concentrations when the mode spot size $S_1$ is kept at 5 and 10 $\mu m$.](image)

![Fig. 14. Sensitivity $\eta_{ab}$ as a function of fluidic gap distance for 5 and 10 $\mu m$ spot sizes.](image)

![Fig. 15. (a) and (b) Plots of the sensitivity $\eta_{ab}$ as a function of mode-mismatch for various spot sizes from two different vantage points.](image)
increases along with the mode spot size. Only 39.5% (or 4 dB power loss) of the input power is also, the ability to detect the smallest changes in concentration improves. However this can happen at the cost of higher power loss of 20 dB and light transmittance was reduced to just 1%.

As pointed out earlier the absorption of optical power is an inevitable process in any integrated optofluidic system. Therefore sensors based on absorbance sensing can be used universally for analyzing all types of fluids. In contrast use of waveguides—a preliminary investigation,” Nature, vol. 442, pp. 381–386, Jul. 2006.

REFERENCES

TABLE II
HEMOGLOBIN CONCENTRATION DETECTION LIMITS FOR VARIOUS MODE SPOT SIZES

<table>
<thead>
<tr>
<th>No.</th>
<th>Spot Size $S_1$ ($\mu m$)</th>
<th>Maximum Sensitivity $n_{ab}$ (Arb.units)</th>
<th>Fluidic Gap Distance $d_{gap}$ ($\mu m$)</th>
<th>Detection Limit $(g/dl)$</th>
<th>Power Loss (dB)</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>5</td>
<td>0.01994–0.02029</td>
<td>72–90</td>
<td>2.5153 x 10^{-7} to 2.4642 x 10^{-7}</td>
<td>4</td>
</tr>
<tr>
<td>2</td>
<td>10</td>
<td>0.02839–0.02854</td>
<td>114–150</td>
<td>1.7612 x 10^{-7} to 1.7544 x 10^{-7}</td>
<td>5.4</td>
</tr>
<tr>
<td>3</td>
<td>20</td>
<td>0.03375–0.03406</td>
<td>158–210</td>
<td>1.4815 x 10^{-7} to 1.4679 x 10^{-7}</td>
<td>6.5</td>
</tr>
<tr>
<td>4</td>
<td>30</td>
<td>0.03492–0.03515</td>
<td>167–250</td>
<td>1.4318 x 10^{-7} to 1.4224 x 10^{-7}</td>
<td>7.5</td>
</tr>
</tbody>
</table>

Table II shows the maximum sensitivity, ideal fluidic gap distance, and the detection limit (g/dl) corresponding to mode spot size $S_1$ varying between 5 and 30 $\mu m$. The detection limit is calculated from (16) assuming that the PD is capable of detecting at least 5 nW change in power output. A detection limit of 1.4679 x 10^{-7} g/dl (or 0.0272 nM) can be achieved for a mode spot size $S_1$ of 20 $\mu m$, which is the best compromise between the sensitivity and the optical power loss.

**A. Sensor Design for Absorbance-Based Sensing**

When absorbance is the basis for detection, the sensitivity improves along with the mode spot size $S_1$ (see Table II); also, the ability to detect the smallest changes in concentration improves. However this can happen at the cost of higher power loss occurring in the fluidic gap (In contrast, in case of refractive-index-based detection, the sensitivity decreases with increase of the mode spot size $S_1$). For instance, when mode spot size $S_1$ is 5 $\mu m$ and the fluidic gap distance $d_{gap}$ is 90 $\mu m$ the maximum sensitivity is achieved. However only 39.5% (or 4 dB power loss) of the input power is transmitted or coupled to the output waveguide. Again if mode-spot size $S_1$ is increased from 5 to 20 $\mu m$ and fluidic gap distance $d_{gap}$ is also increased from 90 to 160 $\mu m$, sensitivity is further enhanced. But this leads to more power loss (6.5 dB power loss) and transmittance is only 22.3%. Thus increment in sensitivity is accompanied by reduced transmittance of power which will make detectability of optical response increasingly difficult. Our simulation results are in agreement with the experimental results reported by Mogensen et al. [10] who had investigated absorbance detection for 100 and 1000 $\mu m$ path lengths (fluidic gap distance). They have reported higher sensitivity when the path length was increased from 100 to 1000 $\mu m$. But this was achieved with large power loss of 20 dB and light transmittance was reduced to just 1%.

The Table III given below summarizes two sets of design parameters for optimizing the sensitivity for refractive index and absorbance based sensing. It is possible to fine-tune the design parameters for further enhancing the sensitivity of the sensor. This can be achieved by applying appropriate detection technique and using light of suitable wavelength to which the optical properties of the analyte are most responsive.

**VII. Conclusion**

In this work, we have analyzed an optofluidic sensor for refractive-index and absorbance-based sensing. The most important result of this work is that mode-mismatch and optical power loss due to absorption of light have been utilized to determine the appropriate fluid channel gap distance and mode spot size for maximizing the sensitivity of such sensors for both refractive-index and absorbance-based sensing. In our sensor, light interacts with the entire bulk of the analyte, and changes in refractive index and molar concentration are utilized for sensing. Such an approach to sensing can yield very good refractive index and absorbance detection limits. In the sensors based on refractometric detection, reduction in mode-spot size will lead to enhancement of sensitivity. If however, sensor is based on absorbance sensing the sensitivity is improved by enlarging the mode-spot size. However the extent to which sensitivity can be improved is determined by the practicality of fabricating minimum size of light source and the lowest possible limit of detection of optical response. After giving due consideration to the above stated facts, we have shown through our modeling and simulation, that the lowest possible detection limits of refractive index and molar concentration which can be achieved are 10^{-7} RIU and 0.0272 nM respectively.

**REFERENCES**


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