



Article Development of a Temperature-Controlled Optical Planar Waveguide Sensor with Lossy Mode Resonance for Refractive Index Measurement

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Abstract: The generation of lossy mode resonances (LMR) with a metallic oxide film deposited on an optical fiber has attracted the attention of many applications. However, an LMR-based optical fiber sensor is frangible, and therefore it does not allow control of the temperature and is not suited to mass production. This paper aims to develop a temperature-controlled lossy mode resonance (TC-LMR) sensor on an optical planar waveguide with an active temperature control function in which an ITO film is not only used as the LMR resonance but also to provide the heating function to achieve the benefits of compact size and active temperature control. A simple flat model about the heat transfer mechanism is proposed to determine the heating time constant for the applied voltages. The TC-LMR sensor is evaluated experimentally for refractive index measurement using a glycerol solution. The heating temperature functions relative to the controlled voltages for water and glycerol are obtained to verify the performance of the TC-LMR sensor. The TC-LMR sensor is a valuable sensing device that can be used in clinical testing and point of care for programming heating with precise temperature control.

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Copyright: © 2021 by the authors. Licensee MDPI, Basel, Switzerland. This article is an open access article distributed under the terms and conditions of the Creative Commons Attribution (CC BY) license (https:// creativecommons.org/licenses/by/ 4.0/). **Keywords:** heat accumulation; programming heating; heat transfer mechanism for the flat model; polymerase chain reaction; optical planar waveguide; sensitivity

1. Introduction

Thin-film deposited optical sensors have been the subject of many studies and find applications in biosensing technology. Previous studies showed that thin-film overlays that are fabricated onto the core of optical fibers absorb the optical power at the wavelengths at which optical resonance occurs. The type of resonance depends on the dielectric properties of the outer thin film on the optical waveguide. Surface plasmon resonances (SPRs) occur if the real part of the thin-film permittivity is negative and greater in magnitude than both its imaginary part and real parts of the permittivity of the sampled material [1]. The optical spectrum of an SPR sensor is highly sensitive to the surrounding medium. The first SPR-based sensor was proposed in 1982 by Nylander et al. [2]. The sensor consists of an optical prism with a coating of a silver thin film in a Kretschmann configuration. A special layer is immobilized on the metallic layer to capture a biological or chemical analyte and create a biosensor or a chemical sensor [3,4]. However, the sensitivity limits for SPR sensors have been reached [5]. A finite one-dimensional photonic crystal structure named Bloch surface waves (BSWs) has found a large number of applications [6–8] and was proposed as an alternative to SPR [9]. With an electromagnetic wave propagating at the interface between a homogeneous medium, the BSWs can be excited by both s- and p-polarized waves. BSWs offer several advantages compared to SPRs. Their dispersion can be designed at almost any wavelength by properly choosing the refractive index (RI) and thickness of the layers of the photonic crystal. BSW-based sensors are characterized by mechanical and chemical stability, thus permitting operation in aggressive environments [9]. However, the

one-dimensional photonic crystal is a multi-layer stacked structure where a precise coating process is required and that will increase the difficulty for application.

Lossy mode resonances (LMRs) with semiconductor metal oxides as dielectric and lossy materials in the visible band are an important aspect of sensor technology. This involves coupling between dielectric waveguide modes and a lossy mode (a guided mode with a complex effective index) for a semiconductor-clad waveguide, as demonstrated by Marciniack in 1993 [10]. A wavelength-sensitive LMR sensor was demonstrated, and the correlation between the experiment and theory was verified in 2010 [11]. The advantages of LMR sensors are as follows: LMRs undergo excitation for both TM (transverse magnetic) and TE (transverse electric) polarization, generate several resonances in the same spectrum, and allow the position of the resonance in the optical spectrum to be controlled by the thickness of the film [12]. Many other materials can also be used to moderate optical losses; therefore, LMR-based optical fiber sensors are used in physical, chemical, and biomedical sensors [13–17]. LMRs can be generated using a metal oxide thin film, rather than expensive noble metals, such as gold or silver. Optical fiber-based sensors are less costly and more portable than prism-based sensors. They also allow remote detection and are resistant to electromagnetic interference [18,19]. However, optical fiber-based LMR sensors are frangible and difficult to handle during cleaning procedures, coating processes, and surface modification for the detection of a specific analyte. A D-shaped optical fiber [20] for which a portion of the cladding for the fiber is removed is often used to increase sensitivity, but this is not suited to mass production or advanced applications. An optical planar waveguide (OPW) is an alternative to an optical fiber for an LMR-based sensor because it is easier to handle, more robust, and involves less cost [21-23].

Biomolecular interaction analysis (BIA) is very sensitive to the environmental temperature [24,25], and even the measured bias of micro-sensors would be influenced by the ambient temperature variation for screening kits and point-of-care testing. Polymerase chain reaction (PCR) is used in molecular biology, disease diagnosis, agricultural testing, and forensic investigation. PCR is a biochemical process that amplifies a single DNA molecule into millions of copies in a short time. Taq DNA polymerase is an enzyme that is used for PCR. It has a half-life of approximately 40 min at 95 °C [26]. It incorporates nucleotides at a rate of about 60 bases per second at 70 °C and can amplify the lengths of about 5000 bases. The temperature has a significant effect on the mechanism and kinetics of chemical reactions and biomolecular interactions. However, most micro-sensors are usually used to perform regression correction or temperature compensation to correct the measurement results with external temperature sensing [27,28]; even the measurement processes are limited to the laboratory with precise temperature control. At present, the problem of temperature stability for sensors has not been effectively solved. It is worth mentioning that temperature control devices are quite different. The temperature sensor is a passive element with a smaller volume. But the common temperature control device is a relatively larger active component, which requires additional heating electrodes and heating media and is difficult to implement.

This study demonstrates an active temperature-controlled optical planar waveguide sensor with LMR (TC-LMR) for RI measurement. Different external voltages are used to drive the ITO (indium tin oxide) heating electrodes to heat the analyte. In terms of electronics, most ITOs are used as a conductor material, such as conductive electrodes on LCDs (liquid crystal displays) or touch panels. In terms of optics, ITOs are used as a transparent material for the transmission of visible light. The dielectric constant for metal has a large negative real part; therefore, metal is generally used in SPR sensors. The dielectric constant for an ITO has a large positive real part [29], which causes LMR. The OPW is constructed using a dielectric material, and the guiding layer has a greater refractive index than that of the two bounding media to allow total internal reflection for the confined propagation of light. Recently, the authors developed an LMR-based sensor with a planar waveguide structure [30] and demonstrated a novel EF-LMR (electrical field-induced lossy mode resonance) sensor to enhance the effect of LMRs for BSA (bovine

serum albumin) molecule detection [31]. For the TC-LMR sensor, the OPW layer is a glass slab for which one bounding medium is air and the other is an ITO thin film on which an analyte is attached. The TC-LMR sensor consists of a self-lighting alignment mechanism on two sides of the glass slab to allow optical fiber coupling for a complete sensing platform. The heating electrodes are fabricated on an ITO conductive glass plate. If an external voltage is applied to the heating electrodes, an electric current passes through the ITO thin film, and the electrical power is transformed into heat. The TC-LMR sensors are evaluated using glycerol solution and are characterized in terms of the dynamic shift in the LMR wavelength for various externally applied voltages.

Temperature control is a very important parameter for biosensors, but no studies of LMR sensors refer to self-heating or temperature control. This paper aims to develop a TC-LMR sensor with an active temperature control function, in which an ITO film is not only used as the LMR resonance but also as the heating function to achieve the benefits of compact size and active temperature control. The heating temperature relative to the LMR wavelength shift is demonstrated to verify the performance for the temperature-sensitive analyte of glycerol and the temperature-insensitive analyte of water. Stable temperature control could reduce the accumulation of measurement errors caused by temperature disturbances. The TC-LMR sensor could provide a feasible design approach for the development of temperature-sensitive biological sensing technology. To the authors' best knowledge, no previous studies use the voltage to control the temperature of an LMR sensor [32–37].

2. Materials and Methods

2.1. Materials

LMRs were generated by guiding modes in a soda–lime glass slab waveguide that is very transparent to wavelengths of 400 to 800 nm [38] so it can be monitored using less expensive optical light sources and a spectrometer. Glycerol was purchased from Huaho Chemical Co., Taoyuan, Taiwan. This study used glycerol concentrations of 20%, 40%, 60%, 80%, and 100% as the analyte. The aqueous solutions were prepared using ultrapure water (>18.2 MΩ- cm) and a Milli-Q system (Burlington, MA, USA). Some glycerol was stirred and dissolved in water for 10 min at room temperature using an electromagnetic stirrer, to produce an analyte solution. The RI of each glycerol solution was measured using a hand-held refractometer (ATOAGO R5000), which was purchased from ATOAGO Co., Ltd. and has a minimum measurement scale of 0.001 RIU (refractive index unit) for visible light. The RIU scale is ranged by 3-stage switching from 1.333 to 1.520. The analyte was dropped onto a prism, and the RI was measured at the boundary of dark and bright regions using a fine adjustment. The respective RI values are 1.3590, 1.3829, 1.4107, 1.4397, and 1.4707 for glycerol in water solutions with concentrations of 20%, 40%, 60%, 80%, and 100%.

2.2. ITO Thin-Film Coating

Before sputtering, all of the soda–lime glass square sheets (Lifeco Optical Inc., Taichung, Taiwan) with a thickness of 0.4 mm and a side length of 100 mm were pre-cleaned with acetone using purity wipes. The glass sheets were then placed in an aqueous solution of 10% NaOH (Huaho Chemical Co.) at 55 °C for 3 min, rinsed in deionized water for about 1 min, and then checked to ensure that water sheeted off the glass surface immediately. The glass sheets were wiped dry to prevent water spots and were used as the substrate for sputter coating. Sputter coating used equipment that was supplied by IZOVAC Co. using a partial pressure of argon of 9×10^{-2} mbar and an intensity of 150 mA. The resulting station used a gas system based on 2 MFC (Ar and O₂). Each magnetron had a 3-zone gas distribution system with a manual needle valve, in order to allow uniform adjustment, and magnetrons were powered by a 10 kW DC power supply. The base vacuum at the start of the process was 5×10^{-3} Pa. The operating gases were argon and oxygen of 99.99% purity. An ITO target of 99.99% purity was purchased from Solar Applied Materials Technology Co. After sputtering, the ITO glass sheets with 0.4 mm in thickness were cut into 30 mm

squares for use as OPWs of the LMR components. The thickness of the ITO film was measured to be approximately 100 nm by an optical interferometer. Two silver paste bars $(30 \times 5 \times 0.1 \text{ mm})$ were coated on two edges of the ITO film as heating electrodes. The TC-LMR sensor head is $30 \times 30 \times 0.5 \text{ mm}$, where the sensing zoom of $30 \times 20 \text{ mm}$ is the area upon the ITO film without a heating electrode covering.

2.3. LMR Experiment Setup

A typical optical transmission measurement was set up for the LMR spectrum. A halogen white light (AQ4303B, ANDO Co., Ltd., Kanagawa, Japan) with 400~1800 nm wavelength was used as the input source, and an optical spectrometer (AQ6315A, ANDO Co., Ltd., Kanagawa, Japan) was used as the receiver to measure the spectrum for the LMR sensor. A schematic diagram of the details of the experimental setup and a photograph of the TC-LMR sensor with the alignment bulk platform are shown in Figure 1. To allow self-alignment, the bulk platform was designed to accommodate all of the components, and the optical axis was along the center of the OPW. An optical fiber patchcord with an FC connector (M74L01), which has a core/cladding diameter of $400/425 \,\mu\text{m}$ and NA 0.39, purchased from THORLABS, guiding light from a light source was inserted into a hole at the left side of the platform. The other optical fiber patchcord with an SMA connector (QP600-1-UV-VIS), which has a core diameter of 600 µm, purchased from Ocean Optics Co., was inserted into the right side linking to a spectrometer. The cables were fixed on the heating electrodes and were connected to an LPS-305 power supply (USED-EQUIP CO., LTD, Taipei, Taiwan). One end of a K-type temperature sensing wire was attached to the bottom of the glass using heat-resistant tape, and the other end was connected to the thermometer (Victor 6801, Chongqing, China). The measurement system is shown in Figure 1. The ITO material conducts electricity, but it is difficult to achieve a tight connection between the electrical cable and the ITO material. Silver paste was used on the surface of the heating electrodes to firmly attach the cables.



Figure 1. A schematic diagram of the experimental setup and a photograph of the TC-LMR sensor on the optical alignment platform.

3. Results and Discussion

The resistance of the ITO film material was 42.6 Ω . This value was measured between the silver paste electrodes. The analyte was dropped on the sensor, and a measurement for the temperatures and the optical wavelength spectra was taken every 10 s.

3.1. Temperature

To measure the temperature profile, the power supply was used to provide an external voltage, and the temperature of the sensor was increased from room temperature (22.2 °C). The time-dependent increase in temperatures, which is defined as the difference between the system temperature and the ambient temperature, was continuously measured for 200 s

in increments of 10 s. The time–temperature profiles for applied voltages of 1 V, 4 V, and 8 V for glycerol and water are shown in Figure 2. During heating at all driving voltages, the temperature difference increases exponentially with time and approaches a maximum value. A voltage of 8 V yields a respective maximum temperature difference for water and glycerol of 52.6 °C and 94.9 °C. A voltage of 1 V yields a minimum respective temperature for water and glycerol of only about 1.0 °C and 5.1 °C. In general, the temperature required for biosensors such as PCR is 95 °C or 70 °C. Therefore, we focus on the temperature within 100 °C, and the applied voltage is 8 V.



Figure 2. The time courses for temperature difference as a function of heating time using applied voltages of 1 V, 4 V, and 8 V for water and glycerol.

3.1.1. Heat Transfer

The theoretical expression to describe the effect of the heating time on the temperature is derived using the physical parameters for thermal energy generation [39]. The law of conservation of energy states that the power generated in a material is equal to the sum of the power stored in the material and the power dissipated. This is written as

$$P_e = C_p m \frac{dT}{dt} + h_{cr} A (T - T_0)$$
⁽¹⁾

where P_e is the electrical power supplied by the external power supply to the ITO thin film, which generates heat and is calculated using the voltage (*V*) and the resistance (*R*) as $P_e = V^2/R$, C_p is the specific heat capacity, *m* is the mass of the heated material, *A* is the interface area, *T* is the instantaneous temperature, T_0 is the ambient temperature, and h_{cr} is the combined average heat transfer coefficient, which is written as $h_{cr} = h_c + h_r$, where h_c and h_r are the average convection and radiation heat transfer coefficients, respectively.

The first item on the right side of Equation (1) represents the heat generated by the applied voltage, which is conducted to the analyte and increases its temperature. The second term is heat dissipation, which is lost to the surrounding air through convection and radiation. The loss factor and the heat transfer coefficients are determined by fitting the time–temperature profiles to the solution of Equation (1). The cross-sectional area between the electrical leads and the electrodes is very small; therefore, heat transfer through conduction is negligible. The dependent variable *T* is experimentally measured as a function of instantaneous time *t*. The solution to Equation (1) can be written in the form of Equation (2):

$$T = T_0 + \Delta T \left(1 - e^{-\frac{t}{\tau}} \right) \tag{2}$$

where $\Delta T = T_{\infty} - T_0$ is defined as the maximum difference in temperature at $t = \infty$ and t = 0, and τ is the time constant [39] which is defined using Equation (3) as

 τ

$$=\frac{mC_p}{h_{cr}A}\tag{3}$$

The experimental time–temperature data for various driving voltages were fitted to Equation (2), and the fitting curves are plotted in Figure 2. The values of ΔT and τ obtained from the fitting calculation are listed in Table 1.

Table 1. The curve-fitted results for the maximum temperature difference (ΔT) and time constant (τ) for various applied voltages.

Analyte		Water			Glycerol	
Voltage	1 V	4 V	8 V	1 V	4 V	8 V
ΔT (°C)	1.52	24.64	53.89	4.72	36.34	101.12
au (s)	142.86	89.29	50.33	106.38	66.89	38.50
R-squared	0.828	0.925	0.996	0.979	0.996	0.998

The voltage applied to the ITO has the greatest effect on the increase in the maximum temperature difference (ΔT). Sufficient heat energy must be transferred to the analyte. If more thermal energy is absorbed by the analyte than is dissipated to the air, the temperature of the analyte gradually increases. A higher applied voltage produces a greater ramp rate (dT/dt), but the second term on the right side of Equation (1) shows that as the temperature difference increases, the rate of heat dissipation also increases. When the heat generation from the power supplier reaches equilibrium with the heat loss from the system, the temperature has a maximum value of T_{∞} , and the maximum temperature difference ΔT is achieved.

The time constant (τ) is a time interval that is required for the temperature to increase to 63% of ΔT and depends on the mass, the specific heat, the area of the contact interface, and the heat transfer coefficient for the heated material. The experimental results show that the temperature of water increases slowly at a lower voltage (1 V) and has a larger time constant (142.86 s). A voltage increase to 4 V produces a time constant of 89.29 s, and a voltage of 8 V produces a time constant of 50.33 s. The same volume of glycerol and water is used in the experiment. The density of glycerol is 1.26 g/cm³. The mass of glycerol and water at a temperature of 20 °C are 2.380 and 4.182 (kJ/kg K) [40,41]. The specific heat ratio for glycerol to water is 0.56. If the contact area and the thermal conductivity of the analyte to the air are assumed to be the same, the heating time constant ratio of glycerol to water is 0.72. The results in Table 1 show that the time constant ratio is between 0.74 and 0.76, which is the result of division by 106.38/142.86, 66.89/89.29, and 38.50/50.33. These experimental results are in good agreement with the theoretical value.

3.1.2. Applied Voltage

Increasing the applied voltage from 1 to 8 V produces an increase in the thermal equilibrium temperature for water and glycerol, as shown in Figure 3. For an applied voltage of 1 V, the respective temperature difference for glycerol and water is 5.0 °C and 1.2 °C. For an applied voltage of 8 V, the respective temperature difference is 99.6 °C and 52.6 °C. The applied voltage has a linear relationship with the equilibrium temperature. There are two important parameters for temperature control. The first one is the response time constant for the heating process. According to the experimental results in Table 1, a larger applied voltage can greatly reduce the time constant and achieve a rapid temperature rise. The other one is the stable temperature difference. According to the voltage versus the stable temperature difference diagram for water and glycerol in Figure 3, two linear equations with the applied voltage as an independent variable and the temperature difference as a

dependent variable can be obtained, where the slope is related to the specific heat coefficient of the analyte. The temperature control equations as the function of applied voltages (*V*) for water (ΔT_{water}) and glycerol ($\Delta T_{glycerol}$) are, respectively, shown in Equations (4) and (5).

$$\Delta T_{water}(V) = -7.24 + 7.37V \tag{4}$$

$$\Delta T_{glycerol}(V) = -14.60 + 13.80V \tag{5}$$



Figure 3. Thermal equilibrium temperature for water and glycerol as a function of applied voltages.

The temperature and heat transfer mechanism for a TC-LMR sensor can be described by a simple flat model, as shown in Figure 4. The heat source is the ITO film on the left, the thickness of the heated analyte is L_1 , the periphery is the air, and the heating distance is L_2 . The *y*-axis shows the temperature. Room temperature is T_0 , and the highest temperature on the heating surface is T_H . The temperature decreases linearly with the distance from the heating surface. The temperature on the analyte surface is T_S . The respective temperature gradients in the analyte and air are k_1 and k_2 , and the effective temperature gradient is assumed as k_e .



Figure 4. A simple flat model describing the temperature change and heat transfer mechanism for a TC-LMR sensor, where L_1 and L_2 are heating distances; and k_1 , k_2 , and k_e are temperature gradients.

Heat conductivity depends on the characteristic constant of the material. A continuous supply of heat (thermal energy) results in heat accumulation; therefore, the temperature increases. Heat is transferred from a high-temperature body to a low-temperature body, and when the temperature is equalized, energy transfer ceases. When thermal equilibrium is achieved, the energy provided to the analyte and the air is expressed as Equation (6):

$$Q = C_e m_e \Delta T \tag{6}$$

where C_e is the effective specific heat capacity defined as the combination of the specific heat capacity (C_p) for the analyte and the air, $m_e = AL_e\rho_e$ is the effective mass of the combination of analyte and air, $L_e = L_1 + L_2$ is the effective distance of heat transfer defined as the sum of the thickness of the analyte and the heat transfer distance in air, and ρ_e is the effective density of the analyte and the air. The distance over which heat is transferred is proportional to the temperature difference; therefore, the effective distance over which heat is transferred is a linear function of the temperature difference and is defined as the temperature gradient $k_e = \frac{\Delta T}{L_e}$.

The energy that is absorbed by the analyte during heating is the square of the temperature difference, as shown in Equation (7):

$$Q = \frac{C_e A L_e}{k_e} (\Delta T)^2 \tag{7}$$

The energy supplied by the external power supply is proportional to the square of the voltage, as shown in Equation (8):

$$Q = \frac{V^2}{R} \cdot \Delta t - Q' \cong \frac{V^2}{R} \cdot \Delta t \tag{8}$$

where Q' is the energy absorbed by the thermal meter contact and the electrical cable to the power supply and is small enough to be ignored. Therefore, Equations (7) and (8) show that the voltage is proportional to the temperature difference.

In Figure 3, the temperature difference is slightly higher than the fitted line at the applied voltage of 8 V. It is ITO materials that easily recombine into a lattice structure at high temperatures. The electrons easily pass through an environment in which atoms are arranged neatly. Therefore, the greater applied voltage results in a higher temperature, a smaller resistance, more electric power, and a higher thermal equilibrium temperature.

3.2. LMR Spectrum

The TC-LMR sensor shifts the LMR wavelength because there is a slight change in the RI of the analyte. Figure 5 shows the LMR spectrogram. The *x*-axis shows the wavelength of the incident light, and the *y*-axis shows the transmission of the light in the TC-LMR sensor. Transmission is defined as $T = P_1/P_0$, where P_0 is the transmitted spectrum without the analyte, and P_1 is the transmitted spectrum with the analyte. The wavelength-dependent response to variation in the external RI is normalized using the spectrum for a bare sensor without an analyte surrounded by air (P_0). For the spectral response measurements for various external RIs, the TC-LMR sensor was dropped with aqueous solutions of glycerol with various concentrations. Incident light at a specific wavelength produces LMR, and there is a maximum loss. The point at which the curve dips in the spectrum is the LMR wavelength. Figure 5 shows the spectral response for RIs from 1.3590 to 1.4707. An analyte with an RI of 1.3590 results in an LMR wavelength of 871.5 nm. As the RI of the analyte increases, the LMR wavelength increases to 942.5 nm, and transmission decreases from 0.77 to 0.37.



Figure 5. The spectrum for a TC-LMR sensor for glycerol solutions with a refractive index from 1.3590 to 1.4707, in increments of 0.02.

The performance of an LMR sensor is evaluated in terms of its sensitivity, which is defined as the ratio of the change in the LMR wavelength to the change in the RI of the analyte. The RI is plotted as a function of LMR, and the sensitivity is calculated as the gradient between two adjacent data points, as shown in Figure 6. The respective sensitivities are 332.0, 418.4, 437.7, 551.7, and 1064.6 nm/RIU for the RIs of 1.3590, 1.3829, 1.4107, 1.4397, and 1.4707. As the RI of the analyte increases, the sensitivity gradually increases, which agrees with the results of [11]. The sensitivity of LMR sensors has been reported as 2953 and 1617 nm per RIU for an ITO with a thickness of 115 and 220 nm, respectively [11]. Using a D-shaped optical fiber increases the sensitivity to 6009 nm/RIU, and the maximum sensitivity is 8742 nm/RIU for an RI from 1.365 to 1.380 with an ITO layer of 170 nm thickness [42]. For a ZnO thin film, the sensitivity of an LMR sensor was experimentally determined to be around 760 nm/RIU for an RI of 1.433 [43]. Though the TC-LMR sensor in this study is a little less sensitive than some optical fiber LMR sensors, the proposed one has a simple and robust structure with low-cost materials for easy mass production. It is also eminently suited to surface modification for biosensing applications.



Figure 6. The LMR wavelength and the sensitivities for the TC-LMR sensor as a function of the RIs of glycerol solutions.

Since the LMR wavelength is sensitive to the RI of the analyte, the RI equation as a function of the LMR wavelength (λ) is obtained by the polynomial curve fitting with an R-square of 0.999. It is expressed in Equation (9).

$$n(\lambda) = -13.557 + 0.0315\lambda - 1.647 \times 10^{-5}\lambda^2 \tag{9}$$

3.2.1. Temperature for a Shift in the LMR Wavelength

A voltage of 8 V was applied to the TC-LMR sensor to heat the analyte, and the temperature and the LMR wavelength were measured accordingly. The shift in the LMR wavelength is plotted as a function of the difference in the heating temperature, and the linear fit for water and glycerol is shown in Figure 7, where the red line and the black line represent the respective linear fitting of experimental data for water and glycerol. Water and glycerol can be heated to more than 50 °C, but the changes in the LMR wavelength are quite different. As the temperature increases, the LMR wavelength for glycerol gradually decreases from 942.5 to 929.0 nm, which is a shift of 13.5 nm. The LMR wavelength is blue-shifted (wavelength decreases). The shift in the LMR wavelength for water varies by +/-2 nm. The LMR wavelength changes based on the variation in the analyte RI. The RI of water is insensitive to the temperature; therefore, the gradient for linear fitting approaches -0.004. The RI of glycerol has a linear relationship with the temperature. Increasing the temperature of glycerol to 55.6 °C causes a wavelength shift of 13.5 nm, which gives a corresponding temperature sensitivity for the LMR of -0.253 nm/°C. The experimental results show that the temperature of glycerol increases by 11.3 °C and 23.0 °C, which causes a respective shift in the LMR wavelength of -7.85 nm and -4.37 nm. The results of a previous study [44] show that if the temperature of glycerol increases by 10 $^{\circ}$ C and 20 °C, the RI decreases by 0.004 and 0.007. In terms of the sensitivity (1064.6 nm/RIU) of the TC-LMR sensor, there are theoretical shifts in the LMR wavelength of -4.26 nm and -7.45 nm. The experimental data are consistent with the theoretically calculated results.



Figure 7. Shift in the LMR wavelength for water and glycerol as a function of temperature difference, where the red circles and the black squares represent the respective experimental data of water and glycerol.

3.2.2. Current for a Shift in the LMR Wavelength

ITO exhibits high optical penetration in the visible part of the spectrum and is a commonly used conductive material. Theoretically, if a large current flows through ITO, a large number of electrons in the material move. To determine whether the current affects the LMR wavelength, a voltage of 8 V was applied, and the current flowing through the ITO surface was 0.2 A. The LMR wavelength and the temperature of glycerol were recorded

every 10 s. When an equilibrium temperature was achieved, the supply voltage was turned off, and the heating dissipation occurred at room temperature. The LMR wavelength was then recorded continuously until the temperature returned to room temperature. To analyze the heating profile, the LMR wavelength shift and the temperature difference for the time are shown in Figure 8. When a voltage of 8 V is applied, the temperature of the glycerol increases exponentially by 42.6 °C in 530 s. The temperature also decreases exponentially during heat dissipation and approaches ambient temperature in 400 s after the applied voltage is switched off. There is a shift in the LMR wavelength during both the heating and dissipation processes. In Figure 8, it could be found that the heating process is reversible. The LMR resonance wavelength does not change in both heating and dissipation cases. In other words, the temperature variation below 100 °C does not affect the LMR optical properties of ITO. The half of the figure after the 530th second is a heat dissipation phenomenon. It can be expressed by Equation (10), where $\Delta T_{disp.}$ is the amplitude of the temperature falling, and α is the heat dissipation coefficient. A heat dissipation coefficient of 0.0103 is obtained by the regression analysis with R² = 0.999.



$$\Delta T_{disp.} = \Delta T \left(1 - e^{-\alpha t} \right) \tag{10}$$

Figure 8. Shift in the LMR wavelength and temperature difference for the heating and heat dissipation of glycerol, where the red circles and the black squares represent the respective experimental data of temperature difference and LMR wavelength shift.

The process of heating and dissipation shows the effect of the current on the LMR wavelength shift at the same temperature. The relationship between the shift in the LMR wavelength and the temperature difference is shown in Figure 9, where the black squares and the red circles are the respective experimental data for the temperature rising and falling. There is no hysteresis in the curve; therefore, a current of 0.2 A has no significant effect on the RI of ITO. The current passes through the ITO film of 100 mm thickness and 30 mm width, which gives a cross-sectional area of 3.0×10^{-5} (cm²). The free electron density is calculated to be 2.4×10^{25} (cm⁻³). ITO comprises a solution of indium oxide (In_2O_3) and tin oxide (SnO_2) at a ratio of about 9:1 (wt) and is a heavily doped semiconductor. The main component of ITO is In₂O₃, which has a high density of tin donors (10^{20} to 10^{21} cm⁻³). This high carrier density indicates conductivity close to that of a metal. The free carriers in the conduction band are driven by an applied voltage. ITO exhibits high conductivity because electrons are donated to the conduction band. ITO has good electrical conductivity and features a large number of free electrons. If an external voltage is applied, an electric field is generated and free electrons in the material are accelerated by this internal electric field. This movement creates an electric current. However, these free electrons collide with almost immobile positive ions in the material; therefore, the free electrons do not achieve a net acceleration. The kinetic energy for the movement of free electrons due to their acceleration is converted into vibrational kinetic energy for the positive ions, which is then converted into thermal energy. Therefore, the free electrons have an average constant velocity that is called the drift velocity. A large number of electrons drift to give an electron density of 2.4×10^{25} (cm⁻³). However, the mass of electrons is extremely small and the electrons continue to drift through a unit volume. The material density does not change. A large current increases the number of drifting electrons but the overall mass does not change. Therefore, an increase in the current does not affect the LMR wavelength shift.



Figure 9. The correlation between the shift in the LMR wavelength and temperature difference for an increase and decrease in the temperature of glycerol, where the black squares and the red circles represent the respective experimental data of temperature rising and falling.

4. Conclusions

This study is the first to experimentally demonstrate a TC-LMR sensor that uses an ITO-coated OPW and a pair of heating electrodes to provide a temperature control platform with an external voltage. It exhibits active voltage control, high mechanical stability, and a simple fabrication process for mass production. The temperature and heat transfer mechanism for the TC-LMR sensor is described by a simple flat model. The applied voltage has a linear relationship with the equilibrium temperature. The results show that the use of an external voltage significantly increases the performance of an LMR sensor and allows its application for temperature-sensitive biological detection platforms such as clinical testing and point of care for programming heating with precise temperature control.

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