Miniaturized Real-Time Oxygen Detection Systems Integrated with Optical Fiber by Doping Ru-Based Fluorescence Sensors

Hon-Ming Fang¹† Song-Jeng Huang²† Chin-Lung Yang³* Kung-Chia Young⁴* Cheng-Kai Yu⁵

¹Department of Emergency, Yang-Ming Hospital, Chaiyi 600, Taiwan, ROC
²Department of Mechanical Engineering, National Taiwan University of Science and Technology, Taipei 106, Taiwan, ROC
³Department of Electrical Engineering, National Cheng Kung University, Tainan 701, Taiwan, ROC
⁴Department of Mechanical Engineering, National Chung Cheng University, Chaiyi 621, Taiwan, ROC
⁵Department of Mechanical Engineering, National Taiwan University of Science and Technology, Chaiyi 600, Taiwan, ROC

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Abstract

This study presents and implements a portable, real-time, miniaturized oxygen sensor and detection system. A ruthenium (Ru)-based fluorescent compound is used in conjunction with a light-emitting diode (LED) light source and a phase-shift detection system, which has fluorescent-compound-doped optical fibers and optical source excitation and reception, as well as phase detection circuits. The local oxygen concentrations correlate with the fluorescence phase shifts, which can be determined based on three approaches: an oscilloscope instrument, an integrated chip, and an amplitude-compensated detection circuit. The advantages of the proposed system include the non-consumption of fluorescence dye, no required reference substance, and fast interaction and responses, despite only a 1.4 phase difference from the optical oscilloscope measurement system. The proposed portable oxygen sensor has potential to offer a rapid evaluation of oxygen concentration in blood or tissues.

Keywords: Ruthenium (Ru) fluorescence, Oxygen sensor, Optical fiber, Miniaturization, Phase detection

1. Introduction

Oxygen concentration affects organisms in vivo. For instance, hypoxia (a low or insufficient oxygen concentration) may induce cellular changes, including hypoxia-induced glycolysis and vessel formation [1,2]. In contrast, hyperoxia (a high oxygen concentration) may lead to the production of free radicals, and subsequent local oxidative stress and damage. A real-time oxygen sensor can be useful in alleviating the manipulation of biomedical problems. However, most commercially available oxygen sensors are used in industry, mining, factories, and laboratories. An inadequate oxygen concentration can trigger an alert to avoid personal harm or unexpected mechanical operations [3]. Portable oxygen sensor devices are urgently required for medical uses, since they are not fully applicable to biomedical measurements, such as the breath quotient of newborns and infants, as well as oxygen concentration in blood and tissue [4]. Oxygen sensors are designed according to various physical and chemical principles, including principles of optics, complementary metal-oxide semiconductors, superficial sound waves, paramagnetic effects, electric current effects, electric potential effects, and chemical catalysis [4].

Optical detection is attractive due to its ultrashort response time. Conventional optical instruments consist of lenses, reflectors, and prisms, of which the components should be maintained with strict ambient factors and also fixed in a supporting platform. Such systems are thus impractical for portable applications. This study adopts optical fibers as a low-loss transmission media. For in vivo use of an oxygen sensor, a ruthenium (Ru)-based phase-shift fluorescence detection method is used in this study. Phase-shift-based detection of fluorescent compounds incurs less interference from external environmental factors than does luminous-intensity-based detection [4-6]. The Ru compound is not consumed during fluorescence measurements, making it is suitable for repetitive testing [6-10]. This study uses light-emitting diodes (LEDs) as light sources to minimize size, power consumption, and potential photo-bleaching effects. The
proposed on-site sensors based on an optical fiber detection system have the following advantages:

1. A fully integrated oxygen sensor system was constructed to determine local oxygen concentrations.
2. The on-site Ru-based fluorescence oxygen sensors are miniaturized at low cost. The proposed system is highly promising for use in on-site oxygen monitoring in clinical settings.
3. An amplitude-compensated detection circuit was designed to enhance the phase detection accuracy.

2. Materials and methods

2.1 Theories of fluorescence lifetime

Fluorescence lifetime is defined as the average time that molecules remain in an excited state. Once a fluorescent material is excited by an incident light, most of the fluorescence intensity, \( I(t) \), decays exponentially [7]:

\[
I(t) = I_0 \times e^{-t/\tau}
\]  

where \( I_0 \) denotes the fluorescence intensity under anaerobic conditions and \( \tau \) represents the fluorescence lifetime. Time and frequency domain methods can be used to detect fluorescence lifetime. The present study uses a time domain method to determine the time delay related to the phase shift that occurs between the absorption and emission in the sine wave pattern of fluorescent dye.

Light quenching can be divided into dynamic and static. Dynamic quenching refers to the interaction between the excited two molecules of the quenching material and a light-emitting material. The two molecules collide with each other. The quenching efficiency, lifetime of fluorescent molecules, and the concentration of the quenching agent decrease the fluorescence intensity:

\[
dye^* \rightarrow Q + dye + Q^* \text{ or } Q
\]  

where \( K_q \) represents the bimolecular quenching rate constant. \( Q \) denotes quencher. The oxygen concentration in the fluorescence system is detected based on the intensity decay of the excited fluorescent material. The Stern-Volmer equation is:

\[
I_d/I_0 = 1 + K_{SV} \cdot [O_2]
\]  

where \( K_{SV} \) denotes the quenching constant of Stern-Volmer (\( K_{SV} = K_q\tau_0 \)), 1 and \( I_0 \) represent the fluorescence intensity in an aerobic environment and an anaerobic environment, in relation to fluorescence lifetimes \( \tau \) and \( \tau_0 \), respectively, and \([O_2]\) is the \( O_2 \) concentration. The ambient oxygen \([O_2]\) can be estimated using this equation. Theoretically, a plot of \( I_d/I_0 \) against \([O_2]\) has a linear form with a slope equal to \( K_{SV} \) and an intercept of unity, thus allowing a simple single-point sensor calibration.

Applying Fourier transformation to Eq (1) yields the following phase difference (\( \phi \))-lifetime relationship equation:

\[
\phi = \tan^{-1}\left[ \frac{\text{Im}(\omega)/\text{Re}(\omega)} \right] = \tan^{-1}\left[2\pi\tau/\omega\right]
\]  

Notably, the fluorescence lifetime \( \tau \) can be obtained by incorporating the phase difference into Eq. (4). Then, by using Eq. (3), the ambient oxygen concentration is obtained.

2.2 Fluorescent dye manufacturing

In previous studies, the difference of fluorescent dye lifetime under excitation was applied to detect the oxygen concentration. Most dye compounds should be processed at a high temperature with precise control. An Ru-based fluorescent dye is used for the proposed miniaturized oxygen sensing system. All the manufacturing procedures of the dye are conducted at room temperature. The formulation of porous gel dyes is the same as that shown in [8]. First, 0.689 mL of n-propyltrimethoxysilane (n-propyl-TriMOS), 1.5 mL of 3, 3, 3-trifluoropropyltrimethoxysilane (TFP-TriMOS), and 1.5 mL of ethyl alcohol (EtOH) were mixed, and then sonicated with ultrasound for 10 min. Deionized water (0.633 mL) was then added into the gel, which was further sonicated for 1 min. Then, 0.1 M hydrogen chloride (HCl) (0.08 mL) was added into the gel, which was further sonicated for 1 min. Finally, 0.1 M HCl (0.08 mL) was added into the gel, which was further sonicated for 60 min, to manufacture a porous gel solution (A).

The fluorescent dye manufacturing process is described below [8]. First, Ru(dpp)_3Cl_3·5H_2O [tris(4,7-diphenyl-1, 10-phenanthroline)-ruthenium dichloride], used as the fluorescent dye, was mixed with EtOH into 1 mL of fluorescent dye solution (B). The mixture was stirred for 10 min with a magnetic stirrer. The porous gel solution (A) and fluorescent dye solution (B) were mixed in a ratio of 4:1 to create the fluorescent dye. The compound was stirred for 20 min with a magnetic stirrer, and fluorophore-doped sol (C) was obtained.

Before the fluorescent dye was doped into the optical fiber, an oxygen sensing film was created. First, surface impurities of the sheet glass were removed using acid and alkaline etching solutions. Oxygen-sensitive fluorescent dye was applied evenly on the glass by spin coating to produce the oxygen-sensitive dye film. The specimen was kept in a dark room and brought out for use when necessary. The processing of the fluorescent dye compound was completed at room temperature, without any special temperature control or vacuum equipment, to facilitate production compared to existing methods [4,6,7,9].

Manufacturing the optical fiber probe as an oxygen sensor required cutting the optical fiber into a fit length. In addition, 1 cm of the surface coating was removed at the terminal. The optical fiber cross-section was cut and kept flat by using a fiber cleaver. Acid/alkaline etching was then used to remove impurities.

The small square area of the fiber terminal requires only a small amount of fluorescent dye, reducing production costs for large-scale manufacturing. In dye cladding, the dip-coating method, described in [7], was used. Furthermore, by using a stepper motor with fine control, the optical fiber dip-coating optimized rate of 0.25 mm/s was obtained for both descending and ascending.
3. Framework design of oxygen sensor system

3.1 Framework of optical fiber system

The proposed system integrates the Ru-based fluorescent dye mixture, an optical transmission structure, circuits of processing modules, and a phase detection measurement module system. A signal generator (Meteix, MTX-3240) is used to produce a sine wave at the optimal frequency (20 kHz) to drive the LED. The light source was fed into an optical fiber coupler and transmitted to the sensing terminal, where it subsequently interacted with the fluorescent dye. The excited fluorescence of the mixture dye was reflected back from the sensing terminal. The fluorescence signals passed through two filters to filter out external noisy light, and were then detected by photomultiplier tubes (PMTs) which induce currents. The weak electrical current was amplified by a current amplifier, fed to the phase detection circuit, and finally captured and displayed on the oscilloscope. The whole framework is shown in Fig. 1. The analyzer was replaced by the phase detector circuit for phase measurement. By using the selected modulation frequency to drive the light source, the fluorescent material produced a longer wavelength following irradiation. The fluorescence signal was consequently captured at the same frequency by using a fluorescence detection device. Finally, the phase offset was measured for various oxygenic concentrations. Phase calibration was conducted to mitigate the delay errors caused by the additional detection circuits.

![Figure 1. Measurement setup.](image)

3.2 Optimal modulation frequency

The phase measurement system requires an excitation source driven under a selected reference frequency. The optimal modulation frequency increased system sensitivity. The optimal modulation frequency equation is shown below [10].

To find the optimal modulation frequency, the phase shift variation was defined and differentiated to obtain its maximal value.

\[
\Delta \Phi = \arctan (\omega \tau_1) - \arctan (\omega \tau_2)
\]  

where \(\Delta \Phi\) is the phase shift, and \(\tau_1\) and \(\tau_2\) are the fluorescence lifetimes at two oxygen concentrations. \(\tau_{\text{max}}\) and \(\tau_{\text{min}}\) are the maximum and minimum fluorescence lifetimes that occur at the anaerobic (0% oxygen) and oxygen-rich (100% oxygen) environments, respectively.

\[
\frac{d(\Delta \Phi)}{d\omega} = \left[ \tau_1 - 2 \tau_2 - \tau_1 \right] / \left[ \left( 1 + \omega^2 \tau_1^2 \right) \left( 1 + \omega^2 \tau_2^2 \right) \right] = 0
\]

Therefore, the optimal modulation frequency is:

\[
f_{\text{opt}} = \frac{1}{2\pi} \sqrt{1/\left( \tau_{\text{max}}\tau_{\text{min}} \right)}
\]

For Ru as the oxygen-sensitive fluorescent material, the optimal modulated frequency was found to be 20 kHz [10] after measuring \(\tau_1(\tau_{\text{max}})\) and \(\tau_2(\tau_{\text{min}})\).

3.3 Phase detection circuitry design

Phase detection of the fluorescence through the fiber media is important because it affects the reliability and sensitivity of the proposed sensor. In the phase detection circuit, all large optical devices and detectors were replaced with miniaturized electronic components. After light is received by the photodiode (Hamamatsu, s6430-01) that generates the current, the detected current is converted into a voltage signal by a current-to-voltage converter. Advanced design of a phase detection circuit can overcome the limited detection capability of weak signals from the interference and noise to obtain phase information. During fluorescence detection, the fluorescence signal is extremely weak and susceptible to external interference and system noise. Therefore, the signals are amplified and then filtered by band-reject filters, band-pass filters, and a phase processing circuit, as shown in Fig. 2. Though the reference was strong and simply pure sinusoid waves were present at the selected modulation frequency, this signal was unnecessary for processing by amplifiers or filters. However, the small delay between the measured signals and reference signals from the circuit contributed significantly to the phase measurement. Therefore, a balanced structure was used to reduce the systematic phase errors from the circuits. Other mismatches from the components, including the op-amp and discrete passive components, were assumed to be small, which contributed to measurement error.

![Figure 2. Circuit design chart.](image)

Because of the weak fluorescence signal, the detected signals from the photodiode were amplified with a high-gain low-noise amplifier. Selecting the first-stage op-amp, whose offset current \(I_{\text{offset}}\) was low such as the junction field-effect transistor (JFET)-based first stage was further used two sets of amplifiers, as shown in the circuit layout in Fig. 2. The weak signals were amplified by the first op-amp in inverting configuration to avoid saturation caused by the offset. Simultaneously, the current signals were transformed into
voltage signals in the first stage. The signals were then amplified into a measurable range by the second-stage amplifier. Additionally, the signal was recovered to the original phase after two inverting-configuration amplifiers. The two amplifier circuits avoided the signal from the saturation and subsequent distortion in the first stage. A second-order band-pass filter was added to suppress undesired interference. The external power, which includes 60-Hz AC spur noise, created a significant amount of interference, distorting the weak signals. Therefore, a 60-Hz band-rejection filter [11] was applied to remove the 60-Hz component.

Fluorescence signals are received by a light detection device and transformed into a current signal. After passing through the amplifier and signal processing circuit filter, the output voltage signals are detected by a phase detector circuit to determine the phase difference between the reference signal and the excited fluorescence signal. This circuit was measured using an oscilloscope for time-domain comparison. The AD8302 chip (multiplication function) was used for the phase detection of the returned fluorescence signals and reference signals.

The AD8302 chip detects the phase difference between the two signals. However, the amplitude change reduces the phase detection accuracy. A circuit was added to transform the sine wave into a square wave to balance the strength amplitude of the reference and the detected signals. Finally, the original sinusoidal signals were extracted back to the fundamental frequency by a band-pass filter.

### 3.4 Circuit for amplitude compensation

An amplitude-compensated design was developed to overcome the amplitude imbalance problem in AD8302. The AD8302 chip measures and analyzes the amplitude and phase difference by transforming sine waves into square waves to eliminate differences in the strength of the two amplitudes. However, current noise accompanies the signal and produces phase offset errors. The proposed amplitude-compensated phase detection circuit eliminates the effect of amplitude imbalance on the accuracy of phase changes, as shown in Fig. 3. A coupled path was created to estimate the amplitudes of the measured and reference signals. An inverter circuit was then added to normalize the imbalanced amplitudes. Finally, a multiplier-based phase detection circuit was implemented after a low-pass filter to remove second-order frequency harmonics. Hence, the output improved the phase estimated without a bias because of coupling of the amplitudes.

### 4. Experimental framework and phase measurement

#### 4.1 Evaluating experimental framework

This study integrated an optical fiber system and detection circuits for oxygen sensing based on a room-temperature fluorescent dye. The experimental framework used a gas release system, a fiber coupler, a sub-miniature version A (SMA) connector, a light source, a gas mixing chamber, a Spectrometer (Ocean Optics, USB 4000), and a computer for analysis, as shown in Fig. 1. The terminal of the optical fiber sensor was clad with the fluorescent dye. Through the connection between the SMA adapter and optical fiber coupler, the light source (blue LED) generated a specific-wavelength light into the optical fiber coupler. The incident light was transmitted to the sensing area of the terminal of the optical fiber, producing fluorescence in the fluorescent dye. Finally, the excited fluorescence returned with a specific wavelength in the signal analysis system. The reflected incident light source and the fluorescence signal were transmitted in the same optical fiber in the measurement framework through the fiber coupler. Fluorescence was generated by the excitation of the incident light, making the light intensity of fluorescence significantly smaller than the incident light intensity, which interfered or even blocked the information of the fluorescence. A customized setup of the corresponding narrow band-pass filters (NBPF) was required to remove noise and interference to avoid phase interference or light overlap of the two light sources.

#### 4.2 Framework of fluorescence lifetime evaluation

The fluorescence lifetime evaluation involved generating pulse waves using a pulse signal generator (Sony, AFG320) to drive an LED source. The fluorescence lifetime refers to the recession time from the excitation to the recession of initial duration. Therefore, rapidly detecting and recording the fluorescence signal is critical for real-time measurement. The optical receiving device in this experiment was replaced with PMTs (Hamamatsu, R928) to improve the sensitivity of fluorescence detection. The signal was connected to a current amplifier (SRS, SR570). Finally, an oscilloscope (GW Instek, GDS-1102) was used to acquire and display the signals.

#### 4.3 Setup of phase detection circuit test

After the phase detection circuit was implemented, circuit reliability and accuracy were confirmed. A photodiode receiver (Hamamatsu, s6430) was used as a light detector. Two filters (incident light filters [ROCOES, B505] and fluorescence passing-through band-pass filters [ROCOES, BP6070]) were added to the framework for light source filtering to reduce the strong leakage signals from the trigger sources. In this experiment, hybrid monolithic NBPF filters were used for fluorescent light filtering. A band-pass filter (Newport, 620BPF) was added to this system test to reduce the noise of the interference filter. The final signals were acquired by the oscilloscope and compared with those processed by AD8302.
5. Results and Discussion

5.1 Fluorescence lifetime

Figure 4 shows the experimental results of determining the fluorescence intensity based on the fluorescent dye for oxygen concentrations of 0 to 100%. According to this figure, increasing the oxygen concentration decreases fluorescence intensity. The optical fiber fluorescence oxygen sensors can quickly respond to oxygen concentration repetitively.

Figure 5 shows the fluorescence lifetime varying with changes in oxygen concentration. The fluorescence lifetime was significantly diminished when the oxygen concentration ranged from 0 to 22.8%. After the measure was normalized according to Eq. (3), a linear increasing trend was observed, as shown in Fig. 6. However, oxygen concentrations exceeding 29%, as marked in the square zone, did not significantly change and were no longer consistent with the Stern-Volmer equation because of the limitation of the instruments and the components. The short lifetime was beyond the responding time of the LED. The fluorescence lifetime distribution of Ru(dpp)$_3$Cl$_2$ ranges from 100 ns to 6 μs [8]. The response time limit of the LED used in this experiment was approximately 1 μs. Therefore, analysis of the lifetime relationship was not valid for fluorescence lifetimes lower than 1 μs. This preliminary experiment verified that the oxygen concentration affects fluorescence lifetime recession, except that the lifetime of the fluorescence was lower than 1 μs, which is beyond the measurement capability of the experiment instruments. Ultrashort laser pulses were required to synchronize the signal collection accurately to reduce lifetime evaluations. Additionally, this oxygen sensor was stable for several weeks during the experimental period. The photostability was estimated to be at least 4 weeks.

5.2 Results of oxygen concentration and phase variation

Signal phase differences are acquired by two methods. The first method uses an oscilloscope, with the PMTs receiving the fluorescence signal. Once amplified by a current amplifier, the signals are captured and analyzed using an oscilloscope based on zero-crossing or peak-to-peak time delay. This signal is not processed by the circuit, and thus, does not contain the delay circuit noise. Based on the AD8302 chip, in the second method, the signal is transmitted through the current amplifier to a phase detection circuit, as shown in Fig. 7(a). In the figure, the error of the phase difference is approximately 1.4 on average. The phase angles of the two data oscilloscope and AD8302 signal were converted to be presented as a Stern-Volmer plot, as shown in Fig. 7(b). An oscilloscope recorded the relationship between oxygen concentration and fluorescence lifetime. Oscilloscope data and AD8302 signal were converted to phase difference values using the Stern-Volmer equation, as shown in Fig. 7(b). These results demonstrate the feasibility of the proposed system in phase detection.

Figure 7(a) also compares oscilloscope-based measurements and those of the phase detection circuit. The chip solution estimated the phase information accurately. Despite a slight signal distortion due to circuit processing, the processed signal from the phase detection circuit did not overlap completely with the oscilloscope data. This is due to the amplitude imbalances of intensity, resulting in errors of the AD8302 chip in the multiplication process of evaluating phase angles.

The AD8302 chip allows a small, low-cost device for accurate phase detection. Following the reception, amplification, and circuit filtering processes, the fluorescence signal is transmitted to the AD8302 chip for phase detection. The phase angle (phase-shift)-oxygen concentration relationship formula was obtained by applying the angle conversion equation given in the AD8302 datasheet. The signal causing a phase delay after chip gain amplification, explaining why the exact value of the signal output compensated for the delay caused by the circuit. Figure 7(a) shows the final phase value. An increasing oxygen concentration monotonically decreased the phase shift due to decreasing fluorescence...
lifetime. Additionally, a decreasing oxygen concentration increased the phase shift caused by fluorescence lifetime. This observation is consistent with the Stern-Volmer equation and earlier analysis.

![Figure 7. (a) Validation of oscilloscope data and AD8302 signal; (b) Stern-Volmer plot of the oscilloscope data and the AD8302 signal retrieval](image)

The repeated experiments of oxygen adsorption and desorption in this system produced consistent results, which are summerized in Fig. 8. According to the figure, a long time was required from the aerobic to anaerobic conditions. Notably, gradually adjusting the concentration of oxygen sped up the responses compared to those of conventional oxygen sensor systems. The transitional response time can be explained as follows. First, the fluorescent gel thickness affects the diffusion rate of oxygen molecules in the gel [9]. Second, the exchange rate of gas diversion produces a delay time that postpones the immediate response. However, the whole system had a relatively quick response time (less than 5 s) compared to those of other types of sensor, such as the Clark-type oxygen gas sensor [12], direct-current highly sensitive galvanic sensor [13], and silicon hotplates for metal-oxide gas sensor [14], whose response times range from 10 to 90 s.

![Figure 8. Repeatable experiment of this oxygen detection system.](image)

### 5.3 Amplitude compensation circuit

Phase detection using the amplitude compensation circuit is shown in Fig. 3. The phase offset signals pass through a fourth-order low-pass filter, and the delay information is obtained for oxygen concentration measurement after the amplitude imbalance is corrected. The amplitude compensation circuit was verified and calibrated before measurement to obtain precise phase measurement. An adjustable phase front-end circuit was implemented for the amplitude compensation circuit. Different amplitude intensities were set up for Channel #1 and Channel #2 in the experiment to observe the tolerance of the phase output when varying amplitudes of intensity. According to the measured results of varying amplitudes of intensity, the detected phase offsets were invariant with imbalanced amplitudes. However, absolute phase errors occurred between the experimental output value and theoretical phase value, and the error increased with increasing phase difference. This was not a random error, and therefore, it was removed using a linear regression method. The calibration produced the real phase measured from the experimental values $A$ (raw voltage data) by applying Eq. (8), which is an experiment-based calibration formula. Therefore, the change in phase difference could be measured with precision. Furthermore, an additional voltage gain lowered the noise interference caused by the low voltage.

$$\Theta = \cos^{-1}\left[\frac{A + (0.267 \cdot A + 0.004)}{10}\right]$$  \hspace{1cm} (8)

By applying the calibrated phase detection circuit, the phase shift was measured and compared with that of the AD8302-based circuit. Figure 9 compares the phase detections of AD8302 signals and amplitude compensation circuit signals. There are a number of inconsistencies between the values measured by the amplitude compensation circuit and those obtained from the AD8302-based circuit measurements. This inconsistency was caused by the error from the nonlinearity of the amplifier. The amplitude compensation circuit focused mainly on removing the phase detection error resulting from the input amplitude imbalance of AD8302. According to this figure, the output curve of the amplitude compensation circuit was more stable than that of AD8302. This finding demonstrates that the proposed circuit can eliminate the errors and interference of amplitude strength differences to provide accurate phase estimation.
5.4 Potential for clinical applications

Maintaining constant oxygen concentration is essential for patients. Many pathological changes, including ischemia, acidosis, metabolic disturbance, tissue degeneration, and even cancer are associated with insufficient local oxygen supply. A real-time oxygen sensor is a useful tool for monitoring immediate oxygen status in clinical settings. Interference from body fluid immersion, biomaterial contact, oxygen diffusion, and unknown oxygen-like biomaterials in vivo needs to be tested and eliminated. The human body temperature varies in a limited range of 35-40 °C. According to a previous study [15], which provided working temperatures for a fluorescence probe of 25-70 °C, a practical application of the proposed oxygen sensor could perform temperature compensation by using a modified Stern-Volmer model. The proposed portable oxygen sensor has potential to be applied to mini-invasive approaches or connected to devices such as endoscopes or needles to detect tissue oxygen.

6. Conclusion

A real-time Ru-based fluorescence optical fiber oxygen sensor was developed for the rapid, sensitive, efficient, and low-cost detection of oxygen. The blending process for the fluorescent dye compound was completed at room temperature without any temperature control or vacuum equipment. Oxygen concentration was measured using phase-shift methodology. In contrast to amplitude-based detection, the proposed oxygen sensor is insensitive to environmental scattered lights or other luminescent interference. The detection system was validated using an oscilloscope and phase detection circuit. Moreover, a miniaturized detection module was implemented based on the AD8302 chip. The signal was enhanced with high-order filters, amplification, and amplitude imbalance compensation. Measurement results demonstrate that the proposed portable oxygen sensor has potential for biomedical applications. In the future, more realistic sensing experiments will be tested for arbitrary oxygen concentrations.

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