Near-Infrared Optoelectronic Non-Invasive Sensing Technology for Glucose Concentration Monitoring in Liquid Solutions

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Abstract. The accurate and non-invasive monitoring of glucose concentration in aqueous solutions is crucial for various biomedical and industrial applications. Despite advancements, traditional glucose measurement methods remain invasive and require frequent sample collection. This study presents an innovative approach using a near-infrared optoelectronic system to measure glucose concentration in pure water. By exploiting specific absorption wavelengths of glucose (1.5 and 1.9 μ m), a reliable, non-invasive detection method is developed. The study involves spectral analysis, experimental validation using glucose solutions of varying concentrations, and the development of a prototype. The proposed system offers a promising solution for precise, real-time glucose monitoring, overcoming the limitations of traditional methods.

Keywords: Non-invasive glucose monitoring, near-infrared sensing, glucose detection, optoelectronic sensor, infrared absorption, Beer–Lambert law, photodiode-based detection, spectral analysis, diabetes management, continuous glucose monitoring

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1. Introduction

The precise and frequent monitoring of glucose concentration in liquid samples is essential in various fields, including biotechnology, food processing, and chemical manufacturing. Traditional glucose measurement methods—such as electrochemical sensors or laboratory-based enzymatic assays—require invasive procedures, limiting their practicality for continuous monitoring. Glucose exhibits strong absorption peaks in the mid-infrared (MID-IR) region, making it highly suitable for optical detection. However, MID-IR LEDs and photodiodes are expensive, which limits their practical implementation.

In this study, we explore an alternative approach by utilizing near-infrared (NIR) LEDs and photodiodes instead of MID-IR components. This approach aims to develop a more cost-effective and accessible solution for glucose sensing while still leveraging the optical absorption characteristics of glucose in the NIR spectral range. Notably, glucose exhibits a distinct absorption peak in the NIR spectrum at approximately 1.5 μ m, facilitating its optical detection based on characteristic absorbance differences compared to pure water. In contrast, at 1.9 μ m, glucose absorption is negligible, making it suitable as a reference wavelength to account for variations in sample thickness and background absorption. For this reason, we selected 1.5 μ m as the primary detection wavelength due to its high glucose absorbance, while 1.9 μ m was chosen as a reference to improve measurement accuracy. This approach fundamentally relies on the Beer–Lambert law, which relates absorbance to the concentration of glucose and the optical path length [1, 2].

To illustrate these absorption characteristics, Figure 1 presents the absorbance spectra of pure water (solid line) and a glucose solution (dashed line) at physiologically relevant temperatures (approximately 37 °C). The region near 1.5 μ m shows significantly higher absorbance for glucose compared to water, making it the optimal wavelength for glucose detection. Conversely, at around

1.9 μ m, glucose absorbance is negligible relative to pure water; thus, this wavelength is employed as a reference to compensate for variations in sample thickness and background absorption [3].



Fig. 1. Comparison of absorbance spectra for pure water and glucose solution at approximately 37 °C, highlighting key absorption differences around 1.5 μm and 1.9 μm. Adapted from Jensen et al [3].

Glucose solutions with known concentrations were prepared and experimentally analyzed using the assembled prototype sensor. Results were compared with theoretical predictions based on the Beer–Lambert law, validating the sensor's accuracy and reliability. This confirms the potential for precise and non-invasive glucose monitoring, supporting further development and optimization of this optoelectronic sensing method for practical applications in biomedical and industrial fields.

2. The Beer-Lambert law

The Beer–Lambert law is fundamental for optical measurement techniques and describes how light is absorbed as it passes through a medium. It relates absorbance (*A*) to the concentration of the absorbing species (*c*), the optical path length (*d*), and the molar absorptivity (ε). This law is widely applied in optical diagnostics; however, practical implementations must account for specific limitations such as scattering, path-length variability, and non-idealities in real-world samples [4].



Fig. 2. Illustration of the Beer–Lambert–Bouguer law. I_0 , intensity of the incident beam; I, intensity of the beam after transmission through a medium of thickness d; c, concentration; ε , extinction coefficient [4].

This law is fundamental for determining glucose concentration in a solution and is expressed as:

$$A(\lambda) = \varepsilon(\lambda) \cdot c \cdot d \tag{1}$$

In our experiment, the absorbance is determined by measuring the ratio of the incident (I_0) and transmitted (I) light intensities:

$$A(\lambda) = \log(\frac{l_0}{l})$$
(2)

By utilizing this equation, a calibration curve can be established, allowing for the determination of glucose concentration in aqueous solutions non-invasively. Since our system relies on a photodiode that converts optical signals into electrical current, the transmitted intensity can be related to the measured photocurrent (I_p) using the responsivity R of the photodiode:

$$I_p = RI \tag{3}$$

Substituting into the Beer-Lambert equation, the glucose concentration can be rewritten as:

$$c = \frac{1}{\varepsilon d} \log(\frac{I_0}{I_p/R}) \tag{4}$$

where I_p is measured directly from the photodiode, and R is taken from its datasheet. The molar absorptivity values ε for glucose at specific wavelengths, including 1500 nm and 1900 nm, are taken from the work of Amerov et al. [5], ensuring the accuracy of the concentration calculations.

3. Experimental setup and measurement procedure

To achieve precise glucose detection, we selected light-emitting diodes (LEDs) operating at wavelengths of 1500 nm and 1900 nm, corresponding to specific absorption characteristics of glucose discussed previously. The 1500 nm LED (MTE5115C1) was employed as the primary detection wavelength due to its significantly higher glucose absorption relative to water. Conversely, the 1900 nm LED (MTE1901W-WRC) served as the reference wavelength, as glucose absorbance at this wavelength is negligible compared to pure water. Utilizing this reference wavelength enabled effective compensation for variations in sample thickness and background absorption, thus ensuring accuracy and consistency in measurements.



Fig. 3. Spectral response of LED MTE5115C1.

The photodiode used in this study was the MTPD2601N-030, selected for its high responsivity in the 1500–1900 nm range. This photodiode features a low dark current, ensuring an improved signal-to-noise ratio, and provides a broad spectral response that ensures compatibility with both LEDs. It plays a crucial role in converting the transmitted infrared light into an electrical current that can be further analyzed for glucose concentration determination.

For the experimental validation, glucose solutions were prepared in distilled water with concentrations ranging from 0% to 16%, increasing in 4% increments. This stepwise preparation allowed for a detailed calibration curve to be constructed. The tested solutions included pure distilled water (0%) as the baseline, followed by solutions with increasing glucose concentrations of 4%, 8%, 12% and 16% with accuracy of \pm 0.2% concentration. Each solution was contained in standard cuvettes to maintain uniform path lengths, ensuring the consistency of measurements.



Fig. 4. Spectral response of LED MTE1901W-WRC.



Fig. 5. Responsivity of photodiode the MTPD2601N-030.

The experimental setup utilized the NI PXIe-4081 precision digital multimeter configured as an amperemeter, measuring the photocurrent generated by the photodiode. The system also included a fixed photodiode and LED positioning system to maintain alignment, along with a cuvette holder to minimize variations in optical path length. The recorded data exhibited a clear correlation where increasing glucose concentration led to a measurable reduction in photodiode current, validating the theoretical predictions based on the Beer–Lambert law presented in Section 2. The collected measurements were used to generate a calibration curve, which provides a reference for estimating glucose concentration in future non-invasive applications.

This experimental analysis confirms the viability of the proposed optoelectronic system for glucose concentration determination. By leveraging near-infrared absorption characteristics and optimizing LED wavelengths, our system demonstrates potential for accurate, non-invasive glucose monitoring. The results indicate that further refinement and optimization of the hardware and signal processing algorithms could enhance sensitivity and extend the applicability of this method to real-



time glucose monitoring in human blood.

Fig. 6. Experimental setup for glucose measurement with a fixed sample position for measurement consistency.

4. Theoretical calculation of photodiode current

The expected photocurrent at the photodiode is determined based on the optical power emitted by the LEDs, their angular distribution, and the responsivity of the photodiode. The total emitted power is distributed over a solid angle Ω , which is given by:

$$\Omega = 2\pi (1 - \cos \alpha) \tag{5}$$

where α is the half-angle of emission. This relation is derived based on the geometry of a conical beam and describes how the radiant flux is spread in space [6]. The radiant intensity I_{max} , defined as the total power emitted per unit solid angle, is given by:

$$I_{max} = \frac{P}{\Omega} \tag{6}$$

where P is the LED's total power output. This equation is widely used in radiometric calculations to determine the intensity of directional light sources [7]. Since the LEDs illuminate the photodiode from a fixed distance of 5 cm, the irradiance E, representing the power per unit area at the photodiode's position, is determined by the inverse square law:

$$E = \frac{I_{max}}{r^2} \tag{7}$$

The photodiode used in this study (MTPD2601N-030) has a lens with a diameter of 4.65 mm, which collects incoming light and focuses it onto the active area of the photodiode. Assuming that all the light incident on the lens is effectively focused on the detector, the total optical power reaching the photodiode is:

$$P_{photo} = E \cdot A_{lens} \tag{8}$$

where A_{lens} represents the area of the photodiode's lens. The photocurrent I_p is then related to the transmitted light intensity and the Beer-Lambert law, which governs the absorption of light in the

glucose solution:

$$I_p = R \cdot P_{photo} \cdot 10^{-\varepsilon cd} \tag{9}$$

This equation provides a theoretical prediction for the photocurrent as a function of glucose concentration. By comparing the calculated I_p values with experimental results, we can evaluate the accuracy of the optical model and validate the measurement system.

5. Results and discussion

Experimental measurements of photocurrents were systematically compared with theoretical predictions derived from the Beer–Lambert law at two selected wavelengths. The analysis included evaluation of deviations between theoretical and experimental results, along with identification of potential sources of error. To clearly illustrate sensor performance, Table 1 presents a summary of calculated and measured photocurrent values for glucose concentrations ranging from 0% to 16%. The data are presented for both wavelengths, facilitating direct comparison and highlighting the sensitivity of the primary detection wavelength (1545 nm) and the stability of the reference wavelength (1900 nm).

Wavelength	1545nm		1900nm	
Method	Calculated (µA)	Experimental* (µA)	Calculated (µA)	Experimental* (µA)
Empty Cell	-	75	-	39.4
Pure Water (0%)	40.81	72.5	5.76	37.5
4%	40.51	71	5.743	36.2
8%	40.21	70	5.727	35.2
12%	39.91	69.3	5.711	34.4
16%	39.62	68.4	5.696	33.9
Errors	± 0.047	± 0.01	± 0.0045	± 0.005

*The current values were obtained by subtracting the dark current of the photodiode from the measured current.

The graphical analysis further illustrates the relationship between glucose concentration and measured photocurrent. Two sets of graphs are provided:

5.1.1545 nm Wavelength

The first set of plots represents the photocurrent behavior at 1545 nm, where glucose absorption is significant. The left panel shows the theoretically calculated photocurrent, while the right panel displays the experimentally measured values. A linear trend is observed in both cases, confirming that glucose absorption follows the expected pattern. However, slight deviations between experimental and theoretical results suggest the presence of external factors such as LED intensity fluctuations, system noise, and potential misalignments in the optical setup.

5.2. 1900 nm Wavelength

The second set of plots corresponds to the 1900 nm reference wavelength, which is used to compensate for variations in optical path length and background absorption. As expected, the measured photocurrent exhibits a lower sensitivity to glucose concentration compared to 1545 nm,

reinforcing its role as a reference signal. The comparison between theoretical and experimental data at this wavelength further validates the system's stability and correction methodology.



Fig. 7. Calculated and experimental photocurrent vs. glucose concentration at 1545 nm.



Fig. 8. Calculated and experimental photocurrent vs. glucose concentration at 1900 nm.

Experimental measurements of photocurrents were systematically compared with theoretical predictions derived from the Beer–Lambert law at two selected wavelengths. The analysis revealed that the measured experimental current values differ from the theoretically calculated ones. This discrepancy arises due to simplifications in the theoretical model, which does not account for certain experimental conditions such as optical misalignment, scattering effects, LED intensity variations, discrepancies in photodiode dark current compensation from datasheet values, and electronic noise in the circuit.

To address this, we introduce an experimental calibration parameter k, which describes experiment configuration.

$$I_p = k \cdot R \cdot P_{photo} \cdot 10^{-\varepsilon cd} \tag{10}$$

This parameter corrects for discrepancies by incorporating factors specific to the experimental setup. After applying this correction, we observe that the theoretical and experimental values align significantly better.

For the 1545 nm wavelength, the calibration parameter was determined as k = 1.74. This

wavelength was chosen because glucose absorption is significantly higher at 1545 nm compared to water, making it the most effective for glucose detection. In contrast, the 1900 nm wavelength was not considered practical for direct glucose measurement, as glucose absorbance at this wavelength is minimal, making it unsuitable for precise concentration determination.

However, it is important to note that the k parameter need to be measured for all different experimental configuration to ensure accurate calibration across various setups. Since different configurations may introduce unique optical and electronic variations, determining the calibration factor individually for each setup is crucial for reproducibility and precision.



Fig. 9. Comparison of theoretical and experimental photocurrent values at 1545 nm after applying the experimental calibration parameter.

With the application of the calibration parameter, the theoretical and experimental data exhibit strong agreement across most concentration levels. However, deviations remain at low and high glucose concentrations, where measurement errors and nonlinear effects become more pronounced. In the intermediate concentration range, the data fit particularly well, confirming the reliability of the model after calibration.

6. Conclusions

This study explored the feasibility of using near-infrared (NIR) optoelectronic components to develop a cost-effective and accessible glucose sensing system. While glucose strongly absorbs in the mid-infrared (MID-IR) region, the high cost of MID-IR LEDs and photodiodes pose challenges for practical applications. To overcome these limitations, we designed a sensing system that leverages NIR LEDs at 1.5 μ m and 1.9 μ m, enabling glucose detection with widely available and lower-cost components.

By applying the Beer–Lambert law, we established a relationship between glucose concentration and the measured photocurrent, allowing quantitative analysis. Experimental validation was conducted using glucose solutions with known concentrations, and measurements were taken using a prototype sensor. The 1.5 μ m wavelength was selected for glucose detection due to its strong absorption, while 1.9 μ m was used as a reference, where glucose absorption is minimal, ensuring a reliable correction for variations in optical path length and background absorption.

A comparison of theoretical predictions and experimental results revealed discrepancies due

to optical misalignment, scattering effects, LED intensity fluctuations, deviations in photodiode dark current from datasheet values, and electronic noise in the circuit. To improve accuracy, an experimental calibration parameter, which describes experiment configuration, was introduced. The calibration factor k = 1.74 was determined for 1545 nm (1.5 µm), significantly improving the agreement between theoretical and experimental data. However, since different experimental setups introduce unique variations, the k parameter need to be measured for all different experimental configuration to ensure consistent performance across various implementations.

After calibration, the sensor demonstrated good agreement between theoretical and experimental data in the intermediate glucose concentration range, validating the effectiveness of the approach. However, at low and high concentrations, deviations persisted, indicating potential nonlinearity in system response or limitations in detection sensitivity. Addressing these challenges will require further optimization of hardware, improved signal processing techniques, and refined calibration methodologies.

Thus, this research presents a viable and cost-efficient alternative to MID-IR-based glucose monitoring by utilizing NIR optoelectronics. By making glucose sensing more accessible and scalable, this technology could contribute to the advancement of continuous and non-invasive monitoring solutions in healthcare and other industries.

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