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Development and experimental validation of a non-invasive near-infrared spectroscopic sensor system for blood glucose monitoring

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ABSTRACT

Non-invasive blood glucose monitoring remains a major challenge in biomedical sensing due to strong light scattering in biological tissues, physiological variability, and limited signal stability of existing optical methods. Near-infrared (NIR) spectroscopy has attracted significant interest as a promising approach for continuous and painless glucose monitoring; however, many reported systems remain confined to laboratory conditions and lack sufficient experimental validation. In this study, a compact multispectral non-invasive sensor system based on NIR spectroscopy is developed and experimentally validated. A mathematical model of optical absorption in biological tissues, based on the Beer–Lambert law and implemented in the MATLAB/Simulink environment, was used to identify wavelength regions exhibiting favorable sensitivity-stability trade-offs. Based on simulation results, four operating wavelengths (940, 1050, 1200, and 1350 nm) were selected for sensor implementation. The proposed system integrates near-infrared light-emitting diodes, a photodiode with low-noise amplification, an analog-to-digital conversion stage, and a microcontroller-based data acquisition unit. Experimental validation was performed under both in vitro measurements using aqueous glucose solutions and in vivo measurements conducted on the human earlobe in a transmission configuration. The results demonstrate a strong correlation between optical signal attenuation and glucose concentration ($r > 0.95$), with a relative measurement deviation not exceeding 5% under controlled experimental conditions. The highest sensitivity was observed at 940 nm, while longer wavelengths (1200–1350 nm) provided enhanced signal stability. Digital signal processing enabled noise reduction of approximately 25–30%, improving measurement reproducibility. Overall, the results confirm the feasibility of the proposed multispectral NIR-based sensor as a proof-of-concept platform for non-invasive glucose monitoring and provide a basis for further optimization and extended experimental and preclinical validation studies.

Keywords: non-invasive glucose monitoring, near-infrared spectroscopy, multispectral optical sensing, biomedical sensor, Beer-Lambert law, wearable biosensors.

INTRODUCTION

Diabetes mellitus represents one of the most serious global health challenges of the modern era. According to the World Health Organization, the number of individuals affected by diabetes has already exceeded 500 million worldwide and is projected to reach approximately 1.3 billion by 2050 [1–3]. This rapid growth places a substantial burden on healthcare systems and underscores the urgent need for improved methods of glucose monitoring [4].

Conventional self-monitoring techniques based on invasive procedures, such as skin pricking and capillary blood analysis, provide high analytical accuracy but are associated with patient discomfort, risk of infection, and the continuous consumption of disposable materials [4, 15]. These limitations restrict their suitability for long-term and continuous monitoring, which is particularly important for patients undergoing intensive insulin therapy [12, 15].

In recent years, optical methods-especially near-infrared (NIR) spectroscopy-have been actively investigated as a promising alternative for non-invasive glucose monitoring [5–7]. The primary advantages of NIR spectroscopy include its relatively deep penetration into biological tissues and its sensitivity to changes in tissue optical properties influenced by glucose concentration [5, 6, 12]. NIR-based optical systems have been increasingly explored in biomedical applications, including metabolite monitoring and wearable sensing platforms, as discussed in recent reviews [6, 7, 14].

Several studies have demonstrated that the wavelength range of approximately 900–1350 nm is particularly suitable for tissue interrogation due to a favorable balance between optical penetration depth and signal stability [5, 12, 16, 18]. Furthermore, multispectral approaches combined with advanced data-processing techniques, including machine learning algorithms, have been shown to improve prediction accuracy and measurement robustness [8, 10, 23]. However, despite these advances, many reported systems remain confined to laboratory environments, and challenges related to signal stability, calibration, and reproducibility under real-world conditions persist [4, 9, 15, 22].

The practical significance of research in this area lies in the potential development of compact, low-cost sensor systems that can be integrated into wearable platforms [11, 12, 14, 20]. Such systems could enable continuous, safe, and user-friendly glucose monitoring, thereby improving disease management and reducing the risk of long-term complications associated with diabetes [11, 12, 15]. Therefore, the development and experimental validation of non-invasive NIR-based sensor systems for glucose monitoring remains an important and timely research direction [4, 9, 14, 22].

BACKGROUND

In recent years, non-invasive glucose monitoring has attracted considerable attention in the fields of biomedical engineering and optical technologies [4, 9, 12, 15]. Conventional methods based on skin pricking and blood analysis provide high analytical accuracy; however, they are associated with patient discomfort, risk of infection, and limited suitability for continuous monitoring [4, 15, 22]. These limitations have stimulated active research into optical approaches, particularly near-infrared (NIR) spectroscopy, which enables the estimation of glucose concentration through analysis of tissue optical properties [5–7, 12, 14].

Several studies have critically assessed the potential of NIR spectroscopy for glucose monitoring [4, 9, 15, 22]. For example, the review presented in [4] highlighted the promise of NIR-based techniques while emphasizing key challenges, including calibration complexity, sensitivity to external factors (such as temperature and motion), inter-individual tissue variability, and long-term instability of sensor systems [4, 9, 15]. These factors significantly reduce both measurement accuracy and reproducibility under practical conditions [9, 15, 22].

Experimental investigations reported in [5] demonstrated that the near-infrared spectral region is suitable for analyzing metabolic processes; however, uncertainty remains regarding the selection of optimal spectral windows [5, 9, 16]. In particular, although the 940–1350 nm range provides sufficient optical penetration depth, its sensitivity strongly depends on individual physiological characteristics, which limits the universality of existing sensor designs [12, 15, 16, 18].

Recent reviews of technological developments [6, 13, 14] have reported substantial progress in optical sensor architectures and signal-processing techniques. Nevertheless, most reported prototypes remain restricted to laboratory environments, and comprehensive clinical validation is still lacking [4, 9, 15, 22]. The absence of large-scale in vivo studies remains

one of the main obstacles to practical adoption of non-invasive NIR-based glucose monitoring systems [9, 15, 22].

In parallel, data-driven approaches have been actively explored [8, 10, 23]. For instance, a hybrid deep-learning model for processing NIR data was proposed in [7], demonstrating improved prediction accuracy [7, 23]. However, the limited availability of large, well-annotated clinical datasets constrains the generalizability of such models [9, 15, 23]. Similarly, the work presented in [8] focused on hybrid convolutional and recurrent neural networks for glucose prediction using pre-recorded spectral datasets. While promising from an algorithmic perspective, this study did not address experimental aspects of optical signal acquisition, including sensor design, optoelectronic calibration, or the influence of physiological and environmental factors such as tissue thickness, temperature, and blood perfusion [9, 15, 16, 22].

Theoretical analyses have also examined the fundamental limitations of NIR spectroscopy for glucose monitoring [9, 10, 15]. In particular, the review in [9] discussed issues related to low glucose selectivity, strong light scattering, and physiological variability [9, 16, 18, 25]. However, this work did not provide experimental validation or concrete engineering solutions to mitigate these limitations [9, 15, 22]. Likewise, the study reported in [10] demonstrated the potential of genetic algorithms, spectrum normalization, and Monte Carlo simulations to improve analytical performance, but was limited to computational analysis without validation on a physical prototype or under varying physiological conditions [10, 15, 22].

Overall, despite notable progress in both experimental and computational approaches, several key challenges remain unresolved [4, 9, 15, 22]:

- insufficient accuracy and reproducibility of NIR-based methods under real-world conditions [9, 15, 22];
- strong dependence of measured signals on individual physiological characteristics and external factors [9, 16, 18, 25];
- imited availability of large clinical datasets for robust model training and validation [8, 15, 22];
- relatively high cost and structural complexity of multichannel optical systems [4, 14, 15];
- difficulty in separating glucose-related spectral features from the dominant absorption of water and other biomolecules [9, 16, 18, 25].

These limitations persist due to both objective factors-such as the complexity of biological tissues, the need for large patient cohorts, and high experimental costs-and subjective factors, including the focus of many studies on laboratory-scale prototypes without transition to extended in vivo validation [9, 15, 16, 22, 25]. Systematization of these unresolved issues leads to the formulation of the central problem addressed in this work: at present, there is no accessible, compact NIR spectroscopic system that has been sufficiently validated to provide accurate and stable non-invasive blood glucose measurements under real-world conditions [4, 9, 15, 22].

RESEARCH AIM AND OBJECTIVES

The aim of this study is to develop and experimentally validate a non-invasive sensor system for blood glucose monitoring based on near-infrared (NIR) spectroscopy [5, 12, 14]. The proposed approach is intended to improve the accuracy and stability of glucose measurements while reducing calibration uncertainty, patient discomfort, and infection risks associated with conventional invasive techniques [4, 15, 22].

In addition, the study aims to demonstrate the feasibility of integrating the proposed sensor system into compact, low-cost, and energy-efficient wearable platforms suitable for continuous and long-term glucose monitoring in everyday conditions [11, 12, 14, 20]. Particular emphasis is placed on achieving stable optical signal acquisition and reproducible measurement performance under controlled experimental conditions [9, 15, 22].

To achieve this aim, the following research objectives were formulated and addressed:

- to develop and analyze a mathematical and simulation model of optical radiation propagation and absorption in biological tissues using the Beer–Lambert law implemented in the MATLAB/Simulink environment, enabling the identification of the most informative spectral range for non-invasive glucose detection [17–19, 25];
- to design and implement the architecture of a non-invasive NIR sensor system, including infrared light-emitting diodes (IR LEDs), a photodiode, signal amplification stages, analog-to-digital conversion, and a microcontroller unit, ensuring stable optical signal acquisition and digital processing [11, 20, 24];

- to perform experimental investigations under in vitro and in vivo conditions in order to validate the proposed model, construct calibration curves, and evaluate the functional performance of the developed sensor system through comparison with laboratory-based and commercial invasive glucose meters [5, 12, 15, 21];
- to conduct a preliminary metrological evaluation of the developed sensor system, including assessment of sensitivity, measurement accuracy, reproducibility, and noise reduction efficiency, with reference to commonly accepted accuracy criteria for glucose-monitoring systems [27–30].

MATERIALS AND METHODS

Object and hypothesis of the study

The object of this research is the interaction of near-infrared (NIR) radiation in the wavelength range of 900–1350 nm with human biological tissue, specifically the earlobe, for the purpose of non-invasive estimation of blood glucose concentration [12, 16, 18, 21].

The central hypothesis of the study is that variations in blood glucose concentration lead to measurable changes in the effective optical absorption of biological tissues in the near-infrared spectral range. This relationship can be approximately described using the Beer–Lambert law and experimentally investigated using a sensor system based on near-infrared light-emitting diodes (LEDs) and a photodiode operating in a transmission geometry [17–19, 20, 25].

To formulate the model and conduct experimental validation, the following assumptions were adopted:

- biological tissue is treated as an optically inhomogeneous medium characterized by averaged absorption and scattering parameters [16, 18, 25];
- physiological factors such as skin temperature, tissue hydration, and blood perfusion are assumed to remain approximately constant within a single measurement series [9, 15, 16];
- motion-related artifacts are minimized by using a fixed V-shaped holder that ensures stable positioning of the optical components [21, 22].
- The following simplifications were applied in the modeling and experimental stages:
- averaged absorption coefficients of glucose, water, and major tissue constituents were used, without accounting for inter-individual physiological variability [16, 18, 25];
- the thickness of the investigated earlobe tissue layer was assumed to be constant [21];
- a direct transmission geometry was considered, while multiple scattering effects were not explicitly modeled [17–19, 25].

Equipment and software

The sensor system was implemented using the following hardware components:

- near-infrared LEDs with central wavelengths of 940 nm, 1050 nm, 1200 nm, and 1350 nm (operating current in the range of 10–20 mA) [11, 12, 14, 20];
- an IRL81A photodiode coupled with a transimpedance amplification stage [20, 24];
- an INA333 instrumentation amplifier with a gain of approximately 51 [20, 24];
- an ADS1115 analog-to-digital converter with 16-bit resolution, ±4.096 V input range, and a sampling rate of 128 samples per second [20, 24];
- an Arduino Nano microcontroller (ATmega328P) for data acquisition and preliminary signal processing [11, 20];
- an HC- 06 Bluetooth module for wireless data transmission [11, 20].

The system was powered by a 9V DC supply, with an LM7805 linear voltage regulator providing a stable 5 V supply to all electronic modules [20, 24].

The software environment included:

- MATLAB/Simulink for mathematical modeling, absorption coefficient analysis, and visualization of simulation results [17–19, 25];
- Arduino IDE for programming the microcontroller and implementing basic signal processing routines [11, 20];
- Microsoft Excel and OriginRo for statistical analysis and graphical processing of experimental data [12, 15].

Theoretical methods

The theoretical framework of this study is based on the Beer–Lambert law, which describes the attenuation of optical radiation as it propagates through an absorbing medium [17–19, 25].

In its generalized exponential form, the transmitted optical intensity can be expressed as:

$$I_{out} = I_0 \, exp(-\alpha L) \quad (1)$$

where: I_{out} – the transmitted light intensity detected by the photodiode; I_0 – the incident light intensity emitted by the near-infrared LED; α – the effective absorption coefficient of the tissue, and L is the optical path length within the biological medium.

For convenience in data processing and analysis, the logarithmic form of the Beer–Lambert law was employed:

$$A = -log_{10}\left(\frac{T}{100}\right) \quad (2)$$

where: A is the absorbance and T is the percentage of transmitted optical power.

The mathematical model was implemented in the MATLAB/Simulink environment as a block diagram, enabling evaluation of the effective absorption coefficient as a function of wavelength and glucose concentration under the adopted assumptions and simplifications.

Scattering effects, wavelength-dependent optical pathlength variations, and tissue heterogeneity were not explicitly modeled and may influence absolute accuracy.

Experimental conditions

In vitro experiments were carried out using aqueous glucose solutions with concentrations ranging from 0 to 300 mg/dL [5, 12, 15]. Measurements were performed at four selected wavelengths (940 nm, 1050 nm, 1200 nm, and 1350 nm) [11, 12, 14]. For each concentration level, the optical response of the developed sensor system was recorded under controlled laboratory conditions [5, 15].

The obtained measurements were compared with reference data acquired using a laboratory spectrophotometer in order to evaluate the consistency of the sensor response and to verify the qualitative agreement between the proposed system and standard optical measurement techniques [5, 15, 16]. These experiments were intended to provide preliminary calibration data and to assess the feasibility of the proposed optical approach [5, 12, 15].

All in vivo measurements were performed on adult volunteers with informed consent and in accordance with institutional ethical guidelines and the principles of the Declaration of Helsinki.

For comparative evaluation, the optical measurements obtained with the proposed sensor system were compared with reference values from biochemical laboratory blood analysis and readings from a commercial invasive glucometer (OneTouch Ultra Easy) [12, 15, 27]. These measurements were performed to obtain a preliminary assessment of the correspondence between optical signal variations and blood glucose concentration [5, 12, 15].

To minimize motion-induced artifacts and ensure stable optical coupling, a V-shaped holder was fabricated using 3D printing technology [21, 22]. This holder provided consistent mechanical fixation and accurate optical alignment between the infrared LED and the photodiode, maintaining a stable optical path geometry during measurements [21]. As a result, the influence of mechanical vibrations and subject movement on the recorded signals was significantly reduced [22].

Overall, this section describes the experimental conditions under which the proposed sensor system was evaluated, including in vitro and in vivo measurement configurations, reference comparison methods, and motion-artifact mitigation strategies [5, 12, 15, 21, 22]. These measures were implemented to support the reproducibility and reliability of the presented experimental results within the scope of a proof-of-concept study [9, 15, 22].

RESULTS AND DISCUSSION

Mathematical modeling of NIR-tissue interaction

Based on the Beer–Lambert–Bouguer law, a mathematical model of optical absorption in biological tissue was implemented in the MATLAB/Simulink environment (Figure 1) [17–19, 25]. The model was designed to evaluate the dependence of optical absorbance on wavelength and to support the selection of an appropriate spectral range for non-invasive glucose monitoring [12, 16, 18].

The simulation results demonstrated a monotonic decrease in the absorption coefficient A with increasing wavelength λ in the range from 400 nm to 1350 nm, which corresponds to the well-known behavior of near-infrared radiation in

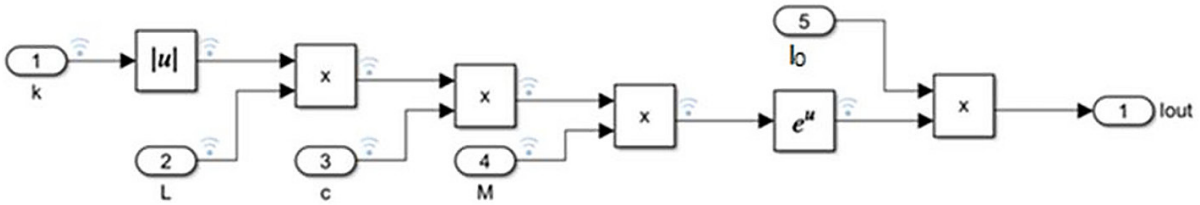

Figure 1. Block diagram of the MATLAB/Simulink model

biological tissues [16, 18, 25]. As the wavelength increased, light penetration depth increased due to reduced scattering and absorption losses [16, 18]. Based on these results, the spectral range from 900 nm to 1350 nm was identified as optimal, providing a balance between sufficient penetration depth and sensitivity to glucose-induced optical changes [12, 14, 16].

Table 1 summarizes the experimental transmission and absorbance data obtained for a set of discrete wavelengths. As the wavelength increased from 400 nm to 1350 nm, the percentage of transmitted radiation increased from 12.8% to 35.4%, while the corresponding absorbance decreased from 0.89 to 0.45. This trend confirms that longer near-infrared wavelengths penetrate deeper into biological tissue [16, 18, 25].

For each wavelength λ_i, the corresponding transmittance value T_i was experimentally measured, and the absorbance A_i was calculated using Equation 2. The dependence $T=f(\lambda)$ was approximated using polynomial regression, yielding a coefficient of determination R^2=0.992, which indicates a strong correlation between transmission and wavelength [16, 18, 25].

Comparative analysis of the obtained spectra revealed the following characteristics:

- the wavelength of 940 nm exhibited the highest sensitivity to changes in glucose concentration [11, 12, 14];
- the range of 1200–1350 nm demonstrated higher signal stability but lower sensitivity [12, 14, 16];
- a multispectral approach combining these wavelength regions provides the most

Table 1. Absorption coefficient A versus wavelength (λ) of infrared radiation

No.	The percentage of optical radiation transmitted through an organic material	Wavelength (nm)	$A = -\log_{10}(T/100)$
1	12.8	400	0.89
2	14.4	450	0.84
3	16	500	0.80
4	17.6	550	0.75
5	19.2	600	0.72
6	20.8	650	0.68
7	22.4	700	0.65
8	24	750	0.62
9	25.6	800	0.59
10	27.2	850	0.57
11	28.8	950	0.54
12	31.5	1000	0.50
13	32.1	1050	0.49
14	32.5	1100	0.49
15	33.0	1150	0.48
16	33.7	1200	0.47
17	34.2	1250	0.46
18	34.9	1300	0.46
19	35.4	1350	0.45

favorable balance between sensitivity and stability [12, 14, 20].

Figure 2 presents a three-dimensional surface plot illustrating the dependence of absorbance $A = -log_{10}\left(\frac{T}{100}\right)$ on wavelength (X-axis), transmittance (Y-axis), and absorbance magnitude (Z-axis and color scale). The surface was constructed using the experimental data summarized in Table 1. Although Equation 2 explicitly relates absorbance only to transmittance, the wavelength λ acts as a parametric variable influencing the measured transmission values [16, 18, 25].

As shown in Figure 2, the absorbance decreases gradually with increasing transmittance, which is consistent with the Beer-Lambert law [17–19, 25].

Two distinct spectral regions can be identified: a sensitive region around 940 nm, where absorbance varies more rapidly, and a stable region between 1200 nm and 1350 nm, where absorbance remains nearly constant [12, 14, 16]. This behavior indicates reduced wavelength dependence in the longer-wavelength near-infrared region, which is advantageous for calibration stability [16, 18].

Based on this analysis, the wavelength ranges with the highest sensitivity (around 940 nm) and the greatest stability (1200–1350 nm) were selected for further sensor calibration and experimental validation [11, 12, 14, 20]. These results formed the basis for selecting the operating wavelengths and developing the sensor architecture described in the subsequent sections.

Design and development of the sensor system architecture

The sensor system was designed as a compact optoelectronic prototype intended for non-invasive blood glucose monitoring using near-infrared (NIR) radiation [11, 12, 14, 20]. The architecture follows a modular design approach and consists of a sequence of functional blocks, as illustrated in Figure 3 [20, 24].

The system includes the following main components:

- IR LED, which generates near-infrared radiation to illuminate the biological tissue under investigation [11, 12, 14, 20];
- photodiode, which detects the transmitted NIR radiation and converts the optical signal into an electrical current [20, 24];
- amplifier, which increases the amplitude of the weak photodiode signal while minimizing electronic noise [20, 24];
- analog-to-digital converter (ADC), which converts the amplified analog signal into a digital representation suitable for processing [20, 24];
- microcontroller, which performs signal acquisition, basic digital processing, system control, and data handling [11, 20];
- Bluetooth module, which enables wireless transmission of processed data to an external device [11, 20];
- power supply module, consisting of a 9V DC source and an LM7805 linear voltage regulator providing a stable 5 V supply to all system components [20, 24].

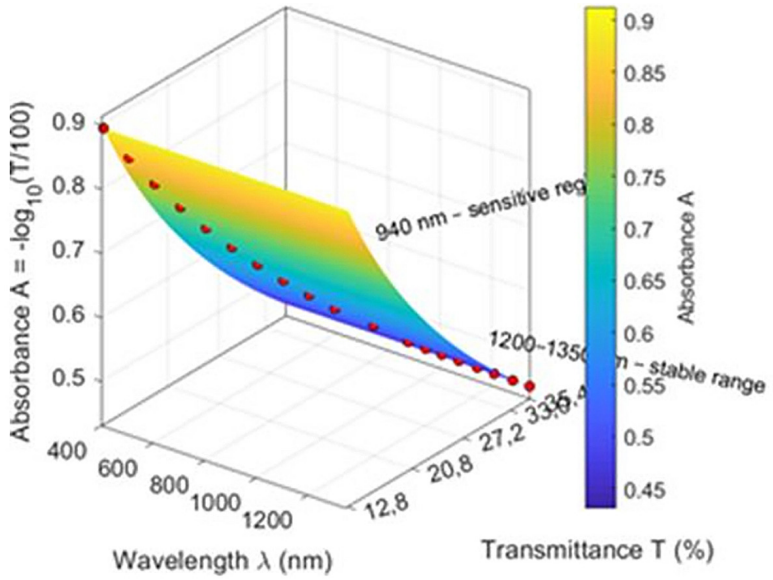

Figure 2. 3D graph, dependence of the absorption coefficient $A = -log10(T/100)$

The software component of the system was implemented using the Arduino IDE, which was used for microcontroller programming, signal acquisition, and preliminary digital filtering [11, 20]. MATLAB/Simulink was employed for advanced signal processing, calibration analysis, and visualization of experimental data [17–19, 25]. The combined hardware–software integration ensured synchronized optical signal acquisition and numerical analysis [20, 24].

As shown in Figure 3, the integration of the optical source, detection, amplification, and data acquisition stages provides a stable signal path from optical emission to digital processing [20, 24]. The modular architecture allows flexible adjustment of operating wavelengths and gain parameters, which facilitates sensitivity optimization and improves measurement reproducibility [14, 20].

Overall, the developed architecture enables reliable optical signal registration with low noise and stable operation [20, 24]. This hardware platform serves as the basis for subsequent in vitro and in vivo experimental validation of the proposed non-invasive glucose monitoring approach [5, 12, 15].

Experimental setup

Experimental validation of the developed NIR sensor system was carried out under both in vitro and in vivo conditions [5, 12, 15]. The objective of these experiments was to verify the proposed mathematical model, construct calibration dependencies, and perform a comparative assessment with laboratory-based and commercial glucose measurement methods [5, 15, 27].

In vitro experiments were conducted using calibration glucose solutions with concentrations ranging from 0 to 300 mg/dL [5, 12, 15]. The optical response of the sensor system was recorded at four operating wavelengths: 940 nm, 1050 nm, 1200 nm, and 1350 nm [11, 12, 14]. For each concentration level, the transmitted optical signal was measured under controlled laboratory conditions [5, 15].

The obtained sensor responses were compared with reference measurements acquired using a laboratory spectrophotometer [5, 16]. These experiments were intended to evaluate the consistency of the optical response and to construct preliminary calibration curves based on the Beer–Lambert law [17–19, 25].

In vivo measurements were performed on human volunteers with the sensor positioned on the earlobe in a transmission geometry [12, 21]. The earlobe was selected due to its relatively uniform thickness and favorable optical properties [21, 22]. Optical measurements obtained with the developed system were compared with reference glucose values obtained from biochemical laboratory blood analysis and a commercial invasive glucometer (OneTouch Ultra Easy) [12, 15, 27].

These measurements provided a preliminary assessment of the correspondence between optical signal attenuation and blood glucose concentration [5, 12, 15]. The observed relative deviation did not exceed approximately 5% under controlled experimental conditions [15, 27].

The experimental setup of the developed NIR sensor system is shown in Figure 4.

The prototype was connected to a personal computer via an Arduino Nano microcontroller, which performed data acquisition and communication using a Bluetooth HC-06 module [11, 20]. The infrared emitter and photodiode were positioned on the volunteer's earlobe using a 3D-printed V-shaped holder, ensuring stable optical alignment and reducing motion-induced artifacts [21, 22].

The electrical signal generated by the photodiode was amplified and digitized using an ADS1115 analog-to-digital converter [20, 24]. The digitized signal was visualized and recorded in real time using MATLAB/Simulink. Measurements were conducted under controlled ambient lighting and temperature conditions [15, 22].

The recorded optical signals were processed to extract quantitative information related to glucose concentration. Signal processing included preliminary filtering, normalization, and calibration based on the Beer-Lambert law [17–19, 25]. Initial signal conditioning was implemented on the Arduino microcontroller, while advanced

Figure 3. Block diagram of the developed sensor system

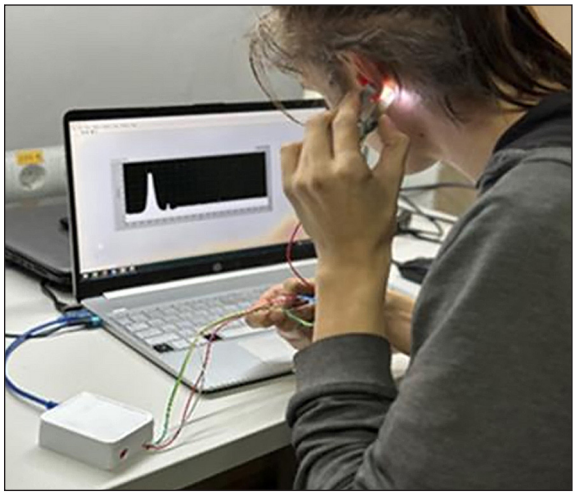

Figure 4. Experimental setup

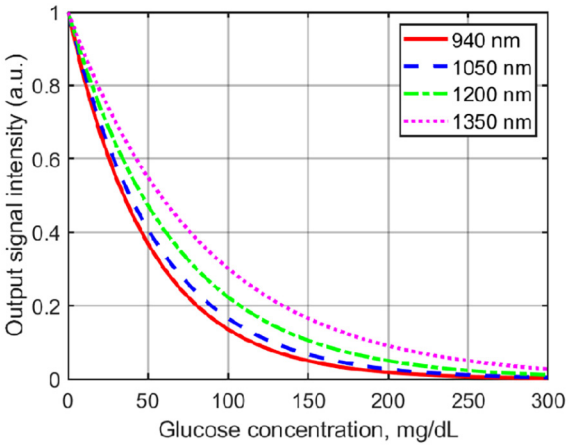

Figure 6. Comparison of sensor response at different wavelengths (940, 1050, 1200, and 1350 nm)

digital filtering, curve fitting, and visualization were performed in MATLAB [20–24].

Calibration curves obtained from in vitro experiments demonstrated an exponential decrease in signal intensity with increasing glucose concentration (Figure 5), consistent with theoretical predictions [5, 12, 17–19].

A strong correlation between optical signal attenuation and glucose concentration was observed (correlation coefficient $r > 0.95$), with a root mean square error RMSE of approximately 0.015 (normalized units) within the investigated concentration range [5, 12, 15].

Figure 6 illustrates the dependence of output signal intensity on glucose concentration for different operating wavelengths. The results confirm that the response at 940 nm exhibits the highest sensitivity to glucose concentration changes, while longer wavelengths (1200–1350 nm) provide more stable but less sensitive responses [11, 12, 14, 16].

Sensitivity analysis, presented in Figure 7, shows that the derivative *dI/dC* reaches its maximum at 940 nm, highlighting the practical importance of this wavelength for non-invasive glucose monitoring [11, 12, 14]. At the same time, the longer-wavelength region offers a stable baseline suitable for calibration and signal normalization [12, 16, 18].

Overall, the experimental results confirm the functional operability of the developed sensor system and demonstrate consistency with the theoretical principles of near-infrared spectroscopy [12, 16–19, 25]. These findings support the feasibility of the proposed multispectral approach for non-invasive glucose monitoring within the scope of a proof-of-concept study [12, 14, 15, 22].

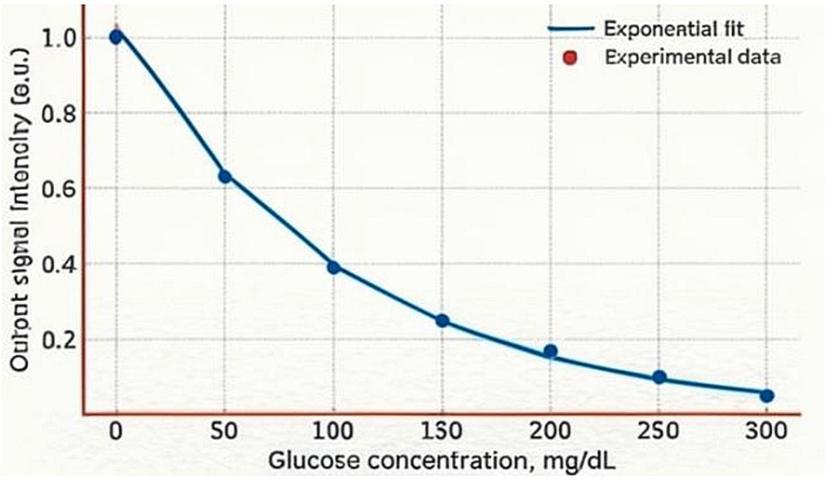

Figure 5. Calibration results and wavelength-dependent sensitivity

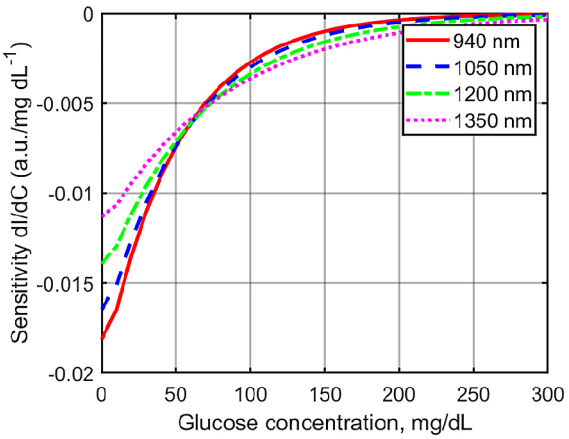

Figure 7. Sensor sensitivity (*dI/dC*) at different wavelengths

Metrological evaluation

Figure 8 illustrates a comparison between the raw (unprocessed) optical signal and the signal after applying digital filtering [20, 24]. As can be observed, the filtered signal exhibits a significantly smoother profile, with a noticeable reduction in the amplitude of random fluctuations compared to the unprocessed signal [9, 15, 22].

The applied digital filtering algorithms resulted in a noise reduction of approximately 25–30%, which is consistent with commonly reported values in optical biomedical sensing systems [9, 15, 22, 24]. Quantitative analysis confirmed that this reduction in noise contributed to improved signal stability, particularly in the presence of motion-related disturbances [21, 22]. As a result, the reliability of optical signal acquisition and subsequent glucose concentration estimation was enhanced [12, 15].

A preliminary assessment of the metrological characteristics of the developed sensor system is summarized in Table 2 [27–30].

The analysis focused on sensitivity, measurement deviation, and signal stability under repeated measurements [27–30]. The sensor demonstrated a sensitivity of approximately 0.0027 units per mg/dL, enabling the detection of relatively small changes in glucose concentration within the investigated range [12, 15].

The observed relative measurement deviation did not exceed approximately 5% under controlled experimental conditions [15, 27]. While these values are within ranges commonly reported in the literature for research-oriented glucose monitoring systems, the present evaluation is intended solely as a preliminary assessment and is provided for comparative reference only, without implying formal compliance or certification with respect to ISO 15197:2013 [27].

Repeated measurements performed under identical conditions did not reveal significant signal drift, indicating satisfactory short-term stability of the developed system [9, 15, 22]. These results suggest that the proposed sensor architecture and signal processing approach provide a stable and reproducible measurement framework suitable for further refinement and extended validation [12, 14, 15, 22].

Overall, the metrological evaluation confirms that the objectives of the study were achieved within the scope of a proof-of-concept investigation and supports the feasibility of the proposed non-invasive NIR-based glucose monitoring approach. [12, 14, 15, 22, 27].

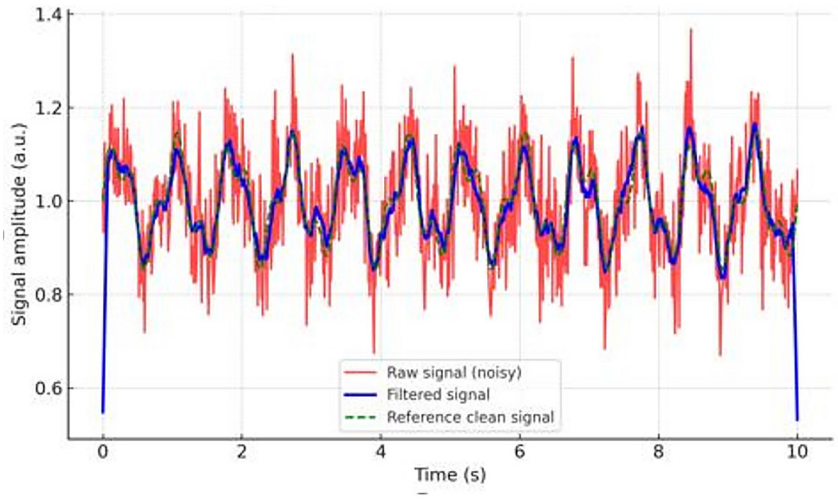

Figure 8. Comparison of raw and filtered signals

Table 2. Preliminary metrological characteristics of the developed NIR sensor system

Characteristic	Value	ISO 15197:2013 Requirement
Sensitivity	~0.0027 units/mg·dL	–
Measurement error	≤ 5%	≤ 5%
Signal stability	No significant drift observed during repeated short-term measurements under controlled conditions	Must be ensured

CONCLUSIONS

This study investigated the feasibility of a non-invasive blood glucose monitoring approach based on near-infrared (NIR) spectroscopy [12, 14]. Mathematical modeling enabled the identification of wavelength regions exhibiting increased sensitivity to glucose-related optical changes, while the developed hardware architecture demonstrated stable optical signal acquisition under controlled conditions [16–19, 25].

Experimental results obtained in both in vitro and in vivo configurations confirmed a measurable correlation between glucose concentration and optical signal attenuation [5, 12, 15, 21]. The application of digital signal processing techniques contributed to improved signal stability and noise reduction, supporting the reproducibility of the recorded measurements [9, 15, 22, 24]. Within the investigated experimental conditions, the observed measurement deviations remained within ranges commonly reported in the literature for research-oriented glucose monitoring systems, without implying formal compliance with clinical standards [15, 27].

Overall, the presented results demonstrate the functional operability of the proposed multispectral NIR sensor system and confirm its suitability as a proof-of-concept platform [12, 14, 22]. The developed approach provides a basis for further optimization, extended experimental validation, and future studies focused on improving robustness, calibration strategies, and long-term stability for non-invasive glucose monitoring applications [14, 15, 22].

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