Available *online* at: http://ajeeet.ft.unand.ac.id/

Andalas Journal of Electrical and Electronic Engineering Technology

ISSN 2777-0079

Advancing Glucose Monitoring: A Near-Infrared Device for Continuous Non-Invasive Assessment

Julius Kirui[1,2], Mathew K Munji[2]. Raphael L Nyenge[2]

[1] *Karatina University, 1957, Karatina, 10101, Kenya*
[2] *Kenyatta University, 43844, Nairobi, 00100, Kenya*

ARTICLE INFORMATION

Received: April 3, 2024
Revised: May 11, 2024
Accepted: July 4, 2024
Available online: July 23, 2024

KEYWORDS

Glucose Monitoring, Non-Invasive Assessment, Device Development, Optical Sensors, Diabetes Management

CORRESPONDENCE

Phone: +254721165964
E-mail: kirui854@gmail.com

A B S T R A C T

Diabetes is a chronic disease that affects millions of people worldwide, with the number of newly diagnosed cases continuing to rise at an alarming rate. Patients diagnosed with diabetes require routine monitoring of their blood glucose levels, which is commonly done through invasive and minimally invasive techniques. However, these methods are often painful, prone to infection, and costly in the long run. Therefore, it is necessary to develop non-invasive, pain-free, affordable glucose monitoring techniques. Near-infrared spectroscopy has emerged as a promising technique among the non-invasive techniques for glucose monitoring, with several studies reporting its efficacy in detecting glucose concentration. This study designed and developed a near-infrared system based on an 1150nm wavelength for continuous non-invasive glucose monitoring. The system gave the output voltages that vary with increased glucose concentrations. It was ascertained from the Clark error grid analysis that the glucose concentration falls within region A and region B. This implies that the device is suitable for measuring glucose concentration non-invasively. As per Bland Altman's analysis, the majority of the data points were within both limits of agreement except one data point that was outside the upper limit of agreement, indicating 93% degree of accuracy, and the residual plot indicates that the regression model for the system is appropriate for predicting non-invasive glucose measurements.

INTRODUCTION

There has been an alarming increase in the number of newly diagnosed diabetic cases across numerous communities globally. This is due to either insufficient insulin release or a body's inability to respond to insulin. The major problems that are associated with the disease are heart failure and blindness, amongst many others [1].

It is estimated that by 2040, 642 million people between the ages of 20 and 79 are expected to have diabetes, according to the WHO's 2015 to 2040 estimate [2]. The estimated annual worldwide death rate is 5 million people. The estimated global spending for disease control is Kshs 74.7 trillion [2].

Kenya, in particular, has a 3.4% prevalence of diabetes and the number of diabetics is expected to reach 500 million by 2044. The average cost of the treatment is Kshs 77,081 per individual. This is about Kshs 38.5 billion [3]. Patients maintain a healthy diet and take insulin to keep their blood sugar levels stable, and routine blood glucose monitoring using invasive and minimally invasive techniques is necessary [4]. Despite being functional, these methods are painful, prone to infections, and expensive [5].

Significant research and development efforts are being made to use a variety of optical approaches to test blood glucose non-invasively. These include Raman Spectroscopy, Fluorescence, Mid Infrared (MIR), Bio-impedance, and Near Infrared (NIR) Spectroscopy. These new technologies present a hope of providing diabetic patients comfort, hope, and relief because they do not require body pricking and are affordable [6]. The quantification of blood glucose is made possible by the variation of light intensity in correlation to transmittance and reflectance. It is established that glucose absorption takes place at 940nm, 1150nm, 1400nm, and 1600nm peak points [7] [8], [9]. In addition, research indicates that compared to other non-invasive techniques, near-infrared (NIR) spectroscopy gives acceptable results, and its accessories are readily available and inexpensive [10]. The application of the NIR method is conceivable since NIR spectroscopy can penetrate the skin up to a depth of 1-100 mm. Water content in blood can be readily avoided by selecting the right wavelength range [4], [11].

As a result of the promising nature of outcomes obtained by using near-infrared techniques to determine glucose non-invasively, numerous studies have been carried out, particularly with the use of near-infrared techniques [12].

https://doi.org/10.25077/ajeeet.v4i1.96

The investigation done by [13] used two distinct, independent circuits in their study to estimate blood glucose with IR spectroscopy. The first circuit is a two-step impedance spectroscopy that operates on the electrical property that indirectly influences changes in human tissue, such as blood glucose level. The second circuit used three NIR wavelengths of 850nm, 940nm, and 1300nm. The system addressed the estimation inaccuracy by comparing the outputs of the two circuits and using an artificial neural network via an external processor. The gadget can also dispense insulin automatically to a diabetic patient who is bedridden in the event of a sudden rise in glucose levels. Notwithstanding the positive outcome, the system is bulky and unadaptable. It is not versatile and exclusively designed for bedridden individuals [13].

In their study, [5] adopted an optical approach. They utilized an LED with a wavelength of 940 nm as a transmitter and a phototransistor as a receiver. The findings from their system produced workable results. However, the technique is vulnerable to noise interference. To eliminate interferences, they suggested using signal conditioning in the circuit [5]. In their system, [14] used wavelengths of 940 nm, 1550 nm, and 1650nm and demonstrated that bilirubin and glucose concentrations are monitored non-invasively.

The hybrid of NIR absorption and Bio-Impedance techniques were utilized by [14]. When compared to the other individual methods, the system produced increased accuracy. This is because of 90% and 10%, established from Clarke Error Grid. These are in the accepted zones A and B, respectively. However, it is subjective because it can only be used by trained individuals and is also sensitive to background light [15].

The study by [8] used 940nm and photodiode and constructed a device for non-invasively measuring patients' blood sugar levels. They obtained promising results and suggested more validation on various data samples for increased accuracy. The research by [16] with IR LED of 980nm and photodiode obtained good results, but the system requires noise filtering and amplification to get accurate results. Also, the use of 1310nm by [10] in their research indicates that red laser light at the band detects glucose non-invasively. The comparative study of [17] using dual IR LEDs of 1300nm and 940nm indicates improved performance, efficiency, and accuracy. However, there is still a challenge of noise interferences and calibration. The investigation by [18] using 860nm and 940nm NIR LEDs posited feasible results but were still affected by noise and calibration issues.

The investigation of [19] and [17] pointed out that indirect measurement, calibration, and systemic noise of non-invasive glucose measuring devices are barrier-inhibiting devices for glucose monitoring non-invasively. On the other hand, the report by [17] shows that there is a high potential for the realization of non-invasive glucose measuring devices because of the promising results obtained, irrespective of the limitations.

The current commercially available devices, which are invasive and minimally invasive despite having benefits, are prone to shorter sensor lifetime, prone to infections, and expensive in the long run [20], [21], [22], [23].

As a result of the challenges encountered, the systems already developed have issues of in-accuracy, noise interferences, and calibration [24], [25], [26], [23]. In addition, some of the developed NIR-based systems use 850nm, 940nm, 1300nm, 1450nm, 1550nm, and 1650nm.

Therefore, from the foregoing discussion, it is apparent that no NIR-based device utilizes a transmitter of 1150 nm, irrespective of the fact that this is one of the peak points for glucose absorption. With this perspective, we propose a precise, non-invasive NIR system utilizing 1150-nm wavelength to ascertain the usability of 1150nm for glucose measurement, eliminate noise, and increase accuracy.

MATERIALS AND METHODS

Methods

The principle of transmittance, as per Beer-Lambert's law, was adopted. The law states that “when a beam of monochromatic light is passed through a solution of an absorbing substance, then the rate of decrease in intensity of radiation with the thickness of absorbing solution is directly proportional to intensity as well as to the concentration of the solution.” [4], [27],[28].

The relationship between absorbance (A) and concentration (C) is illustrated in equations (1) and equation (2).
Therefore, absorbance(A) is given by:

$$A = log\left(\frac{I_0}{I_t}\right) = \varepsilon Cl \quad (1)$$

where;
Ɛ=absorptivity, l=path length, C=concentration
Hence;

$$A \propto C \quad (2)$$

Beer-Lambert’s conditions were observed during the research. The IR LED produced monochromatic light; homogeneous glucose concentration was prepared, and non-electromagnetic materials were used to eliminate interferences. Furthermore, a distance of 1.9cm was maintained between the IR LED and the photodiode. This allows incident radiation to travel an equal and uniform distance in all the samples while testing. Furthermore, the IR light used does not influence the sample’s molecules or atoms, so glucose concentration is not affected by IR light [27].

Testing was done by in-vitro method. The samples for testing were prepared following the table 1. The purpose was to make solutions that fall into all these categories.

Table 1. Glucose Sugar Levels Ranges

Glucose Levels (mmol/L	Glucose Levels (PPM)	Indication
3.8 and below	69 and below	Low
3.9 - 5.5	70 - 99	Normal
5.6 - 6.9	100 - 124	Pre-Diabetic
7.0 and above	125 and above	High

Materials

The materials used are IR LED (L13072-0120P) and InGaAs PIN photodiode (G12180-003A) from Hamamatsu Photonics. The

 https://doi.org/10.25077/ajeeet.v4i1.96

two were used as a transmitter and a receiver, respectively. The IR LED emits infrared light ranging from 1150 to 1250 nm, with a peak sensitivity of 1200nm. It was suitable for the study because 1150nm is one of the peak points of glucose absorption. The photodiode has a spectral response ranging from 900nm to 1700nm. It was selected because it can pick up IR light emitted by the transmitter[29]. The photodiodes are made of silicon, germanium, and lead (II) sulfide. They have spectral ranges of 190-1100nm, 400-1700nm and 1000-3500nm respectively. It was considered a factor because of noise elimination. The low wavelengths of silicon and germanium, which start from 900nm and below, present high noise susceptibility. The photodiodes of lead (II) sulfide have high operating frequency and hence were not the best [19].

An LM358N op-amp, resistors, and capacitors were used for current-to-voltage conversion, filtering circuits, stabilization, and linearization. The reason for selecting LM358N op-amp is that it has a large DC gain, low input offset of 2mV, wide bandwidth, and wide range of operating voltages [30]. The current-to-voltage converter was connected in photovoltaic mode [20] instead of photoconductive mode, which is prone to noise susceptibility. Resistors of 10KΩ and 100KΩ were used to give a gain of 10 to get an appreciable output voltage for analysis. The resistor of 10Ω and 1KΩ was used as coupling and stabilization resistors to the filter's next stage and to prevent the op-amp from going into saturation. A low-pass filter was adopted to attenuate unwanted signals with a cut-off frequency of 1.591KHz. This was achieved using a capacitor of 0.001µF and the determined feedback resistor of 100KΩ.

To get a high linear signal, linearization is essential to prevent the circuit from being noisy and getting into saturation. The circuit was designed with a voltage follower. The reason is to eliminate loading effects from the high filter stage. This is due to the op-amp's high input impedance compared to the circuit's resistances, thus buffering the circuit [22]. The resistor of 100MΩ was used to compensate for voltage changes occurring at the input due to source impedance and bias current and to counteract the voltage offsets as well as high input impedance that occurs during the operation of the system, thus removing loading effects and allowing the same signal transferred to the output [22]. Arduino Mega 2560 was used as a microcontroller. The device was selected because it has many analog and digital pins. Its versatility accommodates many digital inputs and outputs. It has 54 pins for digital input and outputs, 16 for analog inputs, 4 for UART serial ports, and a crystal oscillator. The study aims to coordinate the LCD, the GSM module, and the LED indicators, manipulate analog signals using a linear regression equation fitted in the software, and give output results in mg/dl.

A 3.5’ TFT LCD compatible with Arduino mega 2560 was employed and interfaced through GPIO pins and Atmel studios to display glucose levels in digital format. It was selected for the study because it is a versatile display with a built-in processor and memory for buffering. It can be operated in two modes: 8-bit mode or SPI mode.

A GSM module- SIM800L, was incorporated and interfaced with a microcontroller using Tx/Rx. The AT commands were set to allow tracking of the patient’s status and send results to a person taking care of the patient when there is an increase or decrease of blood glucose above or below the expected level.

The DC-TO-DC buck converter was essential in the project because most of the components were driven by 5V DC except the IR LED, which needed a different voltage. The converter is a step-down device that has a potentiometer of high precision. The device can drive a 3A load without being inefficient. The device can take up an input voltage from 4.5V to 40V and step it down to the required output voltage of between 1.5V to 35V.

LM317 voltage regulator was used to get a biasing voltage for IR LED. It is a three-terminal device capable of providing a voltage output ranging from 1.25V to 37V and a current of 1.5A. The device was set through a voltage divider network of 100Ω and 10Ω to achieve the needed output voltage.

The designed schematic circuit diagram and the implemented system are shown in Figure 1 and Figure 2, respectively.

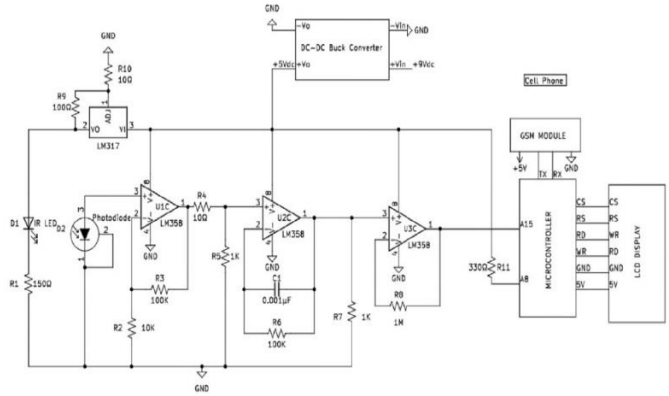


Figure 1: The Schematic Circuit Diagram of the System

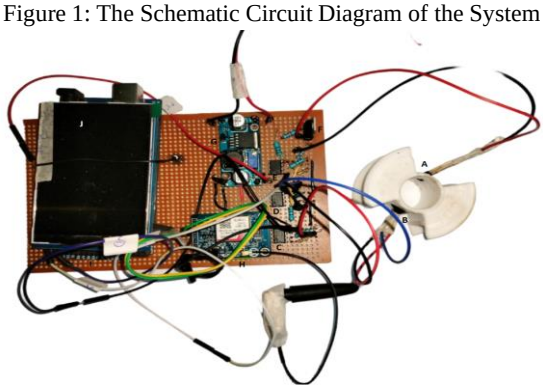


Figure 2: The system implementation on printed circuit board

Key

A – Transmitter (IR LED)
B – Receiver (Photodiode)
C - Current to Voltage Converter
D - Filter
E - Linearizer and Stabilizer
F - LM317 Voltage regulator
G - DC-DC Buck Converter
H - GSM Module (SIM800L)
I – Microcontroller (Arduino Mega 2560)
J - LCD display

RESULTS AND DISCUSSION

Results

The testing was done by using the in-vitro testing method. Fifteen glucose samples were prepared randomly and used for testing. The output voltages were measured, observed, and tabulated as shown in Table 1.

Table 2. Output Voltages for 1150nm

Glucose Concentration (ppm)	Voltage (mV) ± 5
324	2110
270	1850
216	1520
162	1250
126	950
100.8	870
94.68	755
88.56	720
82.44	675
76.32	650
70.2	630
66.6	595
63	556
59.4	530
55.8	515

It was observed that voltage output also increases as glucose concentration increases. This is determined by the amount of glucose concentration present in the samples. It is also evident that there is a linear relationship between the glucose concentration and the voltage outputs. The results of Table 1 are used to plot a scatter plot graph of voltage output against glucose concentration as shown in Figure 3.

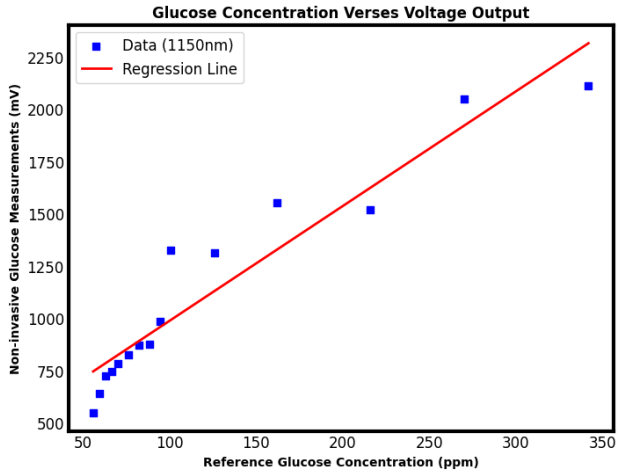


Figure 3. Glucose Concentration Verses Output Voltage

In Figure 3, it is evident that an increase in glucose concentrations gives a corresponding increase in voltage output. Additionally, the output voltage changes in small steps; at high glucose concentrations, the output voltage changes in large steps. The regression equation (3) obtained from the results as a linear fit line was incorporated into the software.

$$y = 5.4820x + 440.33 \quad (3)$$

System Validation

Clarke Error Grid Analysis

The proposed system's efficacy for medical treatment was determined using Clark Error Grid analysis. Figure 4 shows the Clarke Error Grid analysis

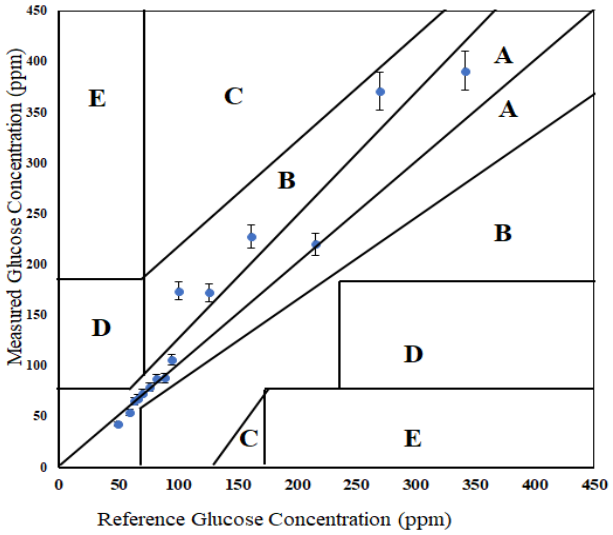


Figure 4. Clarke Error Grid Analysis

The grid has five regions; Region A-20% of values within the reference glucose concentration, hence clinically accurate; B-20% of values are outside the reference glucose concentration, slightly inaccurate but clinically acceptable as it will not lead to inappropriate treatment; C- unnecessary, measurements will lead to overtreatment; D- potentially dangerous, the device fail to detect, and region E-confuse detection of glucose hence will give erroneous results [31],

The results were superimposed onto the Clarke error grid using Excel. According to the Clark Error Grid analysis of Figure 4, 93% of the values fall within region A and 17% within region B. No values are falling within regions C, D, and E. Therefore, the proposed system is suitable for non-invasive monitoring of glucose.

Bland Altman analysis

Bland Altman's plot examines the agreements between two measurement methods of the same variables [32], [33] and are used to determine the degree of accuracy by constructing limits of agreements. The limits of agreement are arrived at as follows: Limits of agreement(upper/lower) = mean of the differences ± 1.96 × standard deviation of the differences. As per [33], Bland Altman recommends that the data points plotted have to fall within the two limits of agreement for the new measurement method to be 95% accurate. Figure 5 shows Bland Altman's plot with data points from the system plotted.

From the observation of Figure 5, most data points fall within the upper limit of agreement and the lower limit of agreement except for one data point that falls outside the upper limit of agreement; this indicates that the proposed device is 93% accurate.

 https://doi.org/10.25077/ajeeet.v4i1.96

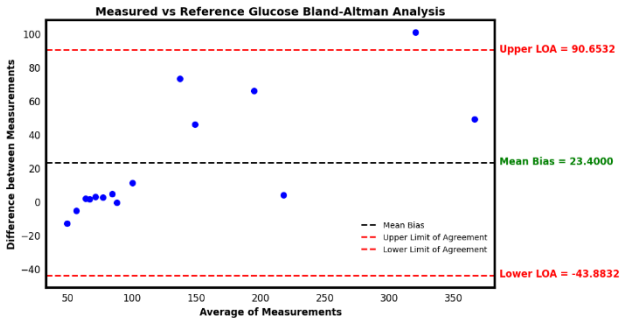
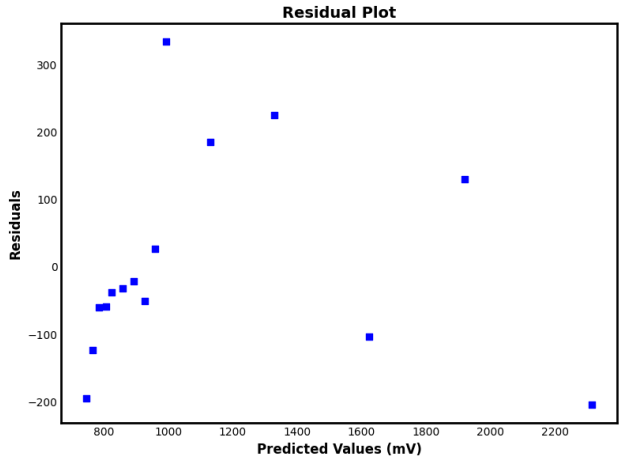


Figure 5. Bland Altman's Analysis

Residual Plot

The random scattering of data points around the zero line in a well-designed residual plot suggests that the residuals are randomly distributed and lack any discernible pattern. The residuals do not capture the underlying relationship between the variables if patterns or trends are observed [34].



The residual plot of the 1150 nm data has no discernible trend, just a random spread of points around the horizontal axis. This shows that the variance of the residuals is essentially constant for various levels of the predictor variable - reference glucose concentrations. As such, it indicates that the homoscedasticity assumption is fulfilled, suggesting that the reference concentrations of glucose do not systematically influence the variation of the non-invasive sugar readings.

Additionally, there is no observable typical pattern from the residual plot, such as funneling or curvature. The residuals' precise random distribution along the horizontal axis indicates a linear relationship between the non-invasive glucose readings and the reference glucose values. Based on the residual plot, the linearity assumption appears to be reasonable.

Moreover, the residual plot lacks any noticeable outliers or pivotal points. Therefore, there appear to be no extreme observations that could adversely affect the regression model. This is because all data points are clustered within a reasonable range around the horizontal axis.

Given this, the residual plot for the data at 1150 nm indicates that the regression model adequately captures the relationship between the non-invasive glucose measurements and reference glucose concentrations. The absence of regular patterns, the random distribution of residuals, and the presence of outliers all suggest that the model satisfies the requirements of linear regression and offers an acceptable fit to the data.

CONCLUSIONS

The results show that 1150nm gives an output corresponding to the glucose concentration applied. Therefore, the system can be used to measure glucose non-invasively. Additionally, when compared with other experimental results of 940nm devices by [17], [18], it is seen that there is a similar relationship in which an increase in glucose solution gives a corresponding increase in voltage output.

As per Clark Error Grid analysis, the system shows values within regions A and B. In Region A, we have 93% of values, and in Region B, we have 17% of values. This agrees with Clark Error Grid since the values for the system meet the 99% criteria stipulated. Given the Bland-Altman analysis, most data points are within the two limits of agreement except for one data point above the upper limit of agreement. This indicates that the system gives a 93% degree of accuracy.

In the analysis of the residual plot, the regression model for the system is appropriate for predicting non-invasive glucose measurements.

LIMITATIONS AND RECOMMENDATION

The scope of the study was limited to in-vitro testing only. Additionally, the transmitter used was 1200nm±50 peak sensitivity.

In our future research, we will use animal subjects, engage large sample sizes, and use commercially available glucose measurement devices to validate the system.

ACKNOWLEDGMENT

We acknowledge the entire Department of Physics of Kenyatta University staff for their immense guidance, support and advice during the entire research period.
The research did not obtain any local or external funding.

REFERENCES

[1] A. Hina, H. Nadeem, and W. Saadeh, "A single LED photoplethysmography-based noninvasive glucose monitoring prototype system," in *Proceedings - IEEE International Symposium on Circuits and Systems*, Institute of Electrical and Electronics Engineers Inc., 2019. doi: 10.1109/ISCAS.2019.8702747.

[2] K. Ogurtsova *et al.*, "IDF Diabetes Atlas: Global estimates for the prevalence of diabetes for 2015 and 2040," *Diabetes Res Clin Pract*, vol. 128, pp. 40–50, Jun. 2017, doi: 10.1016/j.diabres.2017.03.024.

[3] R. Oyando *et al.*, "Patient costs of diabetes mellitus care in public health care facilities in Kenya," *International Journal of Health Planning and Management*, vol. 35, no. 1, pp. 290–308, Jan. 2020, doi: 10.1002/hpm.2905.

[4] S. Haxha and J. Jhoja, "Optical Based Noninvasive Glucose Monitoring Sensor Prototype," *IEEE Photonics J*, vol. 8, no. 6, Dec. 2016, doi: 10.1109/JPHOT.2016.2616491.

[5] P. Narkhede, S. Dhalwar, and B. Karthikeyan, "NIR based non-invasive blood glucose measurement," *Indian J Sci Technol*, vol. 9, no. 41, 2016, doi: 10.17485/ijst/2016/v9i41/98996.

[6] P. S. Reddy and K. Jyostna, "Development of Smart Insulin Device for Non Invasive Blood Glucose Level Monitoring," 2017, doi: 10.1109/IACC.2017.193.

[7] J. Liu, C. Zhu, J. Jiang, and K. Xu, "Scattering-independent glucose absorption measurement using a spectrally resolved reflectance setup with specialized variable source-detector separations," *Biomed Opt Express*, vol. 9, no. 12, p. 5903, Dec. 2018, doi: 10.1364/boe.9.005903.

[8] H. J. Patel, "Near Infrared Spectroscopy: Basic principles and use in tablet evaluation," *International Journal of Chemical and Life Sciences*, vol. 6, no. 2, p. 2006, May 2017, doi: 10.21746/ijcls.2017.2.1.

[9] X. Li and C. Li, "Research on Non-invasive Glucose Concentration Measurement by NIR Transmission," 2015.

[10] K. S. Surendran and T. Sasikala, *Proceedings of the 3rd International Conference on Computing Methodologies and Communication (ICCMC 2019) : 27-29, March 2019*. IEEE, 2019.

[11] M. Vanitha, G. Mathew, K. S. Divya, and R. Vignesh, "Non-Invasive Glucose Estimation Using IR Spectroscopy," *International Journal of Scientific and Research Publications*, vol. 6, no. 5, p. 217, 2016, [Online]. Available: www.ijsrp.org

[12] B. Javid, F. Fotouhi-Ghazvini, and F. Zakeri, "Noninvasive optical diagnostic techniques for mobile blood glucose and bilirubin monitoring," *J Med Signals Sens*, vol. 8, no. 3, p. 125, 2018, doi: 10.4103/jmss.jmss_8_18.

[13] N. D. Nanayakkara, S. C. Munasingha, and G. P. Ruwanpathirana, *MERCon 2018 : 4th International Multidisciplinary Engineering Research Conference : May 30-June 1, 2018, Civil Engineering Complex, University of Moratuwa, Sri Lanka*. 2018.

[14] Aswathy S and Anusree L, "Non-Invasive Glucose Measurement Techniques using Near-Infrared Spectroscopy," 2017. [Online]. Available: www.ijisrt.com268

[15] Vidyadevi M., Chimmad. S., Shilpa, Meghana T., and Tamboti. H, "Non-Invasive Blood Glucose Measurement System," *International Research Journal of Engineering and Technology*, 2018, [Online]. Available: www.irjet.net

[16] I. L. Jernelv, K. Milenko, S. S. Fuglerud, D. R. Hjelme, R. Ellingsen, and A. Aksnes, "A review of optical methods for continuous glucose monitoring," *Applied Spectroscopy Reviews*, vol. 54, no. 7. Taylor and Francis Inc., pp. 543–572, Aug. 09, 2019. doi: 10.1080/05704928.2018.1486324.

[17] Y. J. Heo and S. H. Kim, "Toward long-term implantable glucose biosensors for clinical use," *Applied Sciences (Switzerland)*, vol. 9, no. 10. MDPI AG, May 01, 2019. doi: 10.3390/app9102158.

[18] W. V. Gonzales, A. T. Mobashsher, and A. Abbosh, "The progress of glucose monitoring—A review of invasive to minimally and non-invasive techniques, devices and sensors," *Sensors (Switzerland)*, vol. 19, no. 4. MDPI AG, Feb. 02, 2019. doi: 10.3390/s19040800.

[19] M. A. Al-Dhaheri, N. E. Mekkakia-Maaza, H. Mouhadjer, and A. Lakhdari, "Noninvasive blood glucose monitoring system based on near-infrared method," *International Journal of Electrical and Computer Engineering*, vol. 10, no. 2, pp. 1736–1746, 2020, doi: 10.11591/ijece.v10i2.pp1736-1746.

[20] A. J. Palma, L. Spahić, S. Ilić, M. Anđelković, and G. S. Ristić, "Application of Transimpedance Amplifiers in PIN Photodiode Dosimetry," 2021.

[21] J. Heerlein, M. Behringer, and C. Jäger, "Near-infrared power LED for emerging security and defence applications," in *Electro-Optical Remote Sensing, Photonic Technologies, and Applications V*, SPIE, Oct. 2011, p. 81860O. doi: 10.1117/12.897992.

[22] A. Borel, V. Barzdėnas, and A. Vasjanov, "Linearization as a solution for power amplifier imperfections: A review of methods," *Electronics (Switzerland)*, vol. 10, no. 9. Multidisciplinary Digital Publishing Institute (MDPI), May 01, 2021. doi: 10.3390/electronics10091073.

[23] C. Lorenz, W. Sandoval, and M. Mortellaro, "Interference Assessment of Various Endogenous and Exogenous Substances on the Performance of the Eversense Long-Term Implantable Continuous Glucose Monitoring System," *Diabetes Technol Ther*, vol. 20, no. 5, pp. 344–352, May 2018, doi: 10.1089/dia.2018.0028.

[24] M. Feyikemi Owoeye and A. B. Owoeye, "Implementation of a real-time Arduino Based Non-Invasive Blood Glucose Monitoring System," 2024. [Online]. Available: www.ijaar.org

[25] M. Calero, M. Loza, and D. Benitez, "Towards a Low-Cost Non-Invasive Glucose Sensor Using Near-Infrared Spectroscopy: Preliminary Results," in *2022 IEEE Biennial Congress of Argentina, ARGENCON 2022*, Institute of Electrical and Electronics Engineers Inc., 2022. doi: 10.1109/ARGENCON55245.2022.9939839.

[26] G. Saleh, F. Alkaabi, N. Al-Hajhouj, F. Al-Towailib, and S. Al-Hamza, "Design of non-invasive glucose meter using near-infrared technique," *J Med Eng Technol*, vol. 42, no. 2, pp. 140–147, Feb. 2018, doi: 10.1080/03091902.2018.1439114.

[27] T. G. Mayerhöfer, S. Pahlow, and J. Popp, "The Bouguer-Beer-Lambert Law: Shining Light on the Obscure," *Chemphyschem : a European journal of chemical physics and physical chemistry*, vol. 21, no. 18. 2020. doi: 10.1002/cphc.202000464.

[28] Aryal. S, "Beer-Lambert Law- Definition, Derivation, and Limitations." Accessed: Mar. 28, 2024. [Online]. Available: https://microbenotes.com/beer-lambert-law/

[29] D. Sia, "Design of a Near-Infrared Device for the Study of Glucose Concentration Measurements," 2010.

[30] D. Thakur, K. Sharma, and R. Sharma, "Ultra low-power low-pass filter design for wearable biomedical

applications," in *Proceedings of 4th International Conference on 2021 Devices for Integrated Circuit, DevIC 2021*, Institute of Electrical and Electronics Engineers Inc., May 2021, pp. 629–632. doi: 10.1109/DevIC50843.2021.9455815.

[31] H. Mondal and S. Mondal, "Clarke Error Grid Analysis on Graph Paper and Microsoft Excel," *Journal of Diabetes Science and Technology*, vol. 14, no. 2. SAGE Publications Inc., p. 499, Mar. 01, 2020. doi: 10.1177/1932296819890875.

[32] D. Giavarina, "Understanding Bland Altman analysis," *Biochem Med (Zagreb)*, vol. 25, no. 2, pp. 141–151, 2015, doi: 10.11613/BM.2015.015.

[33] C. S. Hofman, R. J. F. Melis, and A. R. T. Donders, "Adapted Bland-Altman method was used to compare measurement methods with unequal observations per case," *J Clin Epidemiol*, vol. 68, no. 8, pp. 939–943, Aug. 2015, doi: 10.1016/j.jclinepi.2015.02.015.

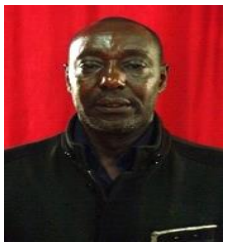
[34] M. Kozak and H. P. Piepho, "What's normal anyway? Residual plots are more telling than significance tests when checking ANOVA assumptions," *J Agron Crop Sci*, vol. 204, no. 1, pp. 86–98, Feb. 2018, doi: 10.1111/jac.12220.

NOMENCLATURE

WHO	World Health Organization
mV	Millivolt
KΩ	Kiloohm
Ω	Ohm
µF	Microfarad
nm	Nanometer
NIR	Near-infrered
LED	Light Emitting Diode
PPM	Parts Per Milliom
Mmol/l	Milimoles Per Litre
GSM	Global System for Mobile Communication
TFT	Thin-Film-Transistor Liquid-Crystal Display
UART	Universal Asynchronous Receiver / Transmitter
GPIO	General Purpose Input/Output
Tx/Rx	Transmit and Receive
AT	ATtention

AUTHOR(S) BIOGRAPHY

JULIUS KIRUI obtained his B.Sc degree in Electronics and Communications from Masinde Muliro University in 2014. He is currently doing an M.Sc. in Electronics and Instrumentation at Kenyatta University, Nairobi, Kenya.



MATHEW K. MUNJI obtained his BSc in Physics degree in 1992, and Mphil in Physics degree from Moi University, Eldoret, Kenya. He did PGDE in 2004 from Egerton University and in 2011 received a Ph.D. in 2011 from Nelson Mandela Metropolitan University, Port Elizabeth, South Africa. Dr. Munji has published 24 papers in refereed Journals and 20 articles from conferences and workshops attended. The current research activities are in Solar Cells, Electronics and Instrumentation.

RAPHAEL L. NYENGE obtained his B.ED (SC) degree in education science from Kenyatta University, Nairobi, Kenya, in 1990 and the M.Sc. degree in Physics from Kenyatta University, Nairobi, Kenya, in 2007, and Ph.D. degree in Material Science at The University of the Free State, South Africa in 2015. He has published 11 papers in refereed journals and 5.