Journal of Electronic Research and Application, 2025, Volume 9, Issue 6
http://ojs.bbwpublisher.com/index.php/JERA
ISSN Online: 2208-3510
ISSN Print: 2208-3502

Development and Characterization of a Non-Enzymatic Glucose Biosensor Based on a Gold Film EG-FET and an Inverting Amplifier Readout Circuit

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Abstract: The development of low-cost, non-enzymatic glucose biosensors is crucial for advancing accessible diabetes management. This paper presents the experimental testing of an extended-gate field-effect transistor (EG-FET) that uses a gold film as the sensing structure. The system innovatively employs a custom-designed inverting operational amplifier circuit for precise signal acquisition and an Arduino Nano platform for real-time data processing and visualization, eliminating the need for expensive laboratory equipment. At the core of the design is a depletion-mode MOSFET, whose current-voltage properties were characterized. The function of the sensor was demonstrated by testing its response to phosphate-buffered saline containing glucose at different concentrations. A clear modulation of the drain current in the linear region of the EG-FET was observed, and a preliminary analysis revealed a linear correlation between the output current and glucose concentration, indicating the system's potential for quantitative detection. This study successfully validates the feasibility of a compact, cost-effective, and non-enzymatic EG-FET biosensing platform, establishing a solid foundation for future development of point-of-care diagnostic devices.

Keywords: EG-FET; Glucose detection; Non-enzymatic biosensor; Readout circuit

Online publication: December 16, 2025

1. Introduction

Biosensors have found extensive applications across a wide range of fields, including medical diagnostics, environmental monitoring, and food safety assessment. Glucose monitoring systems utilized in diabetes management dominate the biosensor market, accounting for approximately 85% of the global market share [1]. The substantial demand for glucose biosensors necessitates the development of simplified, robust detection systems to enhance diabetes management capabilities. Conventional electrochemical glucose biosensors predominantly utilize enzymatic reactions, with glucose oxidase (GOx)-based systems representing the gold standard for clinical and personal monitoring applications. Despite their widespread adoption, these first-generation biosensors present

fundamental limitations, including: (1) finite enzyme stability under physiological conditions, (2) progressive activity loss due to protein denaturation, (3) complex fabrication requirements involving biological immobilization, and (4) elevated manufacturing costs associated with enzyme purification and stabilization [2]. Such constraints significantly impede their adaptation for emerging applications requiring mechanical flexibility, mass-scale production, and continuous monitoring capabilities, thereby driving the development of next-generation non-enzymatic sensing paradigms.

Field-effect transistor (FET)-based sensing platforms have emerged as a transformative approach for next-generation biosensing applications, offering significant advantages in miniaturization, sensitivity, and scalability. The extended-gate field-effect transistor (EG-FET) architecture offers unique advantages through physically separating the sensing interface from the transistor channel [3]. **Figure 1** illustrates the structural configuration of the EG-FET device. In this configuration, an externally extended and functionalized gate terminal interacts with target analytes while shielding the transistor body, significantly improving both operational robustness and design versatility. The detection mechanism relies on surface potential modulation at the extended gate electrode, induced by biochemical binding events or ionic concentration changes, which directly controls the MOSFET drain current, enabling high-sensitivity biochemical detection.

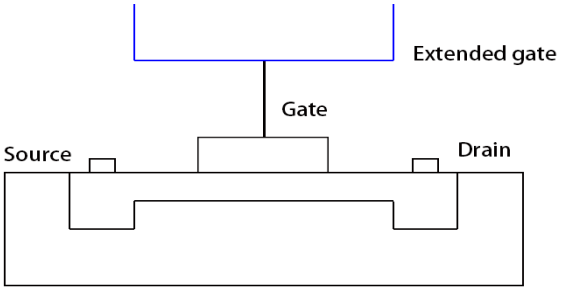

Figure 1. The EG-FET structure

The project's primary objectives focus on designing an inverting operational amplifier-based analog front-end circuit for accurate current–voltage (I–V) characteristic acquisition and implementing a microcontroller interface for real-time data visualization and processing. The extended gate architecture utilizes a gold (Au) thin film that exhibits superior chemical inertness, excellent electrochemical stability, and high electrical conductivity (typically > 4×10^7 S/m) [4]. Artificial glucose solutions prepared in phosphate-buffered saline (PBS) are utilized as testing samples under controlled laboratory conditions to simulate physiological environments. By systematically analyzing the I–V characteristic at different glucose concentrations, the project establishes a calibration curve within the MOSFET's linear operating region, enabling quantitative estimation of glucose levels. The design emphasizes modularity, cost-effectiveness, and scalability, making it suitable for diabetes management.

2. Literature review

2.1. Generational evolution of glucose biosensors

Glucose biosensors have evolved through four distinct generations, with the first three relying on glucose oxidase (GOx) as the biological recognition element. In contrast, the fourth generation comprises non-enzymatic glucose biosensors [5]. The first-generation enzymatic glucose biosensors rely on detecting hydrogen peroxide (H_2O_2)

production or consuming molecular oxygen (O_2), which acts as a natural co-substrate, to determine glucose concentration in the sample. Second-generation enzymatic glucose biosensors employ artificial redox mediators to replace the oxygen-dependent electron transfer mechanism. The third-generation enzymatic glucose biosensors enable direct electron transfer between the enzyme and the electrode, thereby eliminating the need for either natural or synthetic mediators. The fourth generation represents non-enzymatic glucose biosensors utilizing direct electron transfer via nanomaterial-modified electrodes. These sensors operate through glucose molecule adsorption on the electrode surface, followed by concentration quantification through electron flow analysis, thus eliminating dependence on biological recognition elements.

Compared to conventional enzymatic platforms, non-enzymatic glucose biosensors offer three key advantages: (1) enhanced operational stability through elimination of protein components, (2) streamlined manufacturing processes compatible with large-scale production, and (3) extended shelf life - critical characteristics for reliable patient self-monitoring. These attributes directly overcome the inherent weaknesses of enzyme-based systems, namely their susceptibility to environmental degradation and complex biological immobilization requirements.

2.2. The development of FET sensors

The historical development of FET biosensing platforms has followed a clear technological trajectory: beginning with basic MOSFET structures in the 1960s, advancing to ISFETs for ion detection in the 1970s [6], evolving into BioFETs with biological recognition elements in the 1980s–90s, and most recently progressing to EG-FET architectures that physically separate the sensing and transduction functions [3]. This progression reflects an ongoing optimization of sensitivity, specificity, and manufacturability in field-effect biosensing.

The metal-oxide-semiconductor field-effect transistor (MOSFET) is the fundamental building block of modern microelectronics, primarily employed in digital and analog circuits [7]. To adapt FET architectures for chemical sensing applications, the ion-sensitive field-effect transistor (ISFET) was developed by replacing the conventional metal gate with an electrolyte interface, enabling direct detection of ionic species such as pH variations [6]. The sensor platform was significantly advanced through the creation of biologically functionalized FETS (BioFETs), which immobilize specific recognition elements (e.g., glucose oxidase enzymes, IgG antibodies, DNA aptamers) within the gate dielectric region, enabling selective biomolecular detection [8].

The EG-FET offers three fundamental advantages over conventional ISFET and BioFETs designs: (1) Physical separation of the sensing interface from the transistor body prevents electrochemical degradation of the semiconductor channel, significantly enhancing device longevity; (2) Modular gate design enables independent optimization of sensing materials (e.g., Au, graphene) and transistor components, improving both sensitivity and manufacturing yield; (3) Elimination of biological components at the active transistor region circumvents the stability limitations inherent to enzyme- or antibody-based BioFETs.

2.3. Inverting operational amplifier (op-amp) configurations

While inverting operational amplifier (op-amp) measurement circuits have been relatively underexplored in glucose sensor applications, this well-established electronic design technique has been extensively characterized in other contexts. For instance, Ramdeo and Srivastava [9] demonstrated a quad-stage op-amp design utilizing dual-gate MOSFETS. Although their work primarily focused on amplifier architecture, it included detailed schematics and analysis of inverting op-amp configurations that provide valuable insights for MOSFET characterization—

principles that could be adaptively applied to electrochemical biosensing applications.

3. Industrial relevance, real-world applicability, and scientific/societal impact

3.1. Industrial relevance

The growing demand for reliable diabetes management technologies underscores the critical industrial significance of developing high-performance, low-cost glucose sensors. This project employs widely available, mass-produced components such as MOSFETS and op-amps, ensuring minimal production costs. Although conventional enzymatic glucose sensors are widely used, their industrial applicability is constrained by inherent limitations, including enzyme degradation, complex fabrication requirements, and elevated manufacturing costs. In contrast, the proposed EG-FET-based glucose sensor demonstrates high compatibility with standard semiconductor manufacturing processes, facilitating large-scale production. This innovative design addresses the industry's evolving need for durable, scalable, and low-maintenance biosensing technologies, positioning it as an up-and-coming candidate for commercialization in the medical device sector.

3.2. Real-world applicability

The proposed EG-FET glucose sensor is characterized by high integration, compact size, and rapid response, ideally suited for everyday health monitoring applications. Its design primarily addresses the needs of individuals requiring regular glucose tracking, thereby significantly enhancing user convenience. Notably, the sensor exhibits substantial potential beyond glucose detection; altering the sample solution can effectively measure concentrations of various other molecules. The low-cost manufacturing process and a straightforward calibration procedure significantly improve its accessibility for home-based glucose management systems. Consequently, this sensor technology is well-positioned for widespread adoption among diverse economic demographics. Thus, the sensor provides a practical, user-friendly, and versatile pathway toward comprehensive health-monitoring solutions across multiple application scenarios.

3.3. Scientific impact

The scientific innovation of this project lies in implementing an inverting amplifier circuit to bias the MOSFET gate voltage, significantly improving the precision and reliability of current–voltage measurements. Additionally, integrating a microcontroller for data acquisition marks a substantial advancement in the system-level design of biosensors, enabling real-time monitoring and digital signal processing. Compared with conventional ISFET-based sensor platforms, combining a depletion-mode N-channel MOSFET and extended-gate architecture enhances both the detection range and the long-term stability of the device, effectively addressing critical challenges related to sensor lifespan and response reliability.

Moreover, the system's modular and adaptable architecture promotes broader scientific exploration by reconfiguring the sensing interface to detect other biologically relevant analytes. This versatility extends the platform's research impact across multiple sensing domains, including biomedical diagnostics, environmental monitoring, and chemical analysis. As such, the project contributes meaningfully to advancing low-cost, high-precision, and scalable biosensor technologies.

3.4. Societal impact

The EG-FET glucose biosensor holds substantial potential for societal impact by improving access to reliable

health-monitoring technologies. Diabetes, a chronic disease affecting hundreds of millions worldwide, imposes a particularly severe burden on populations in low-resource and underserved regions. By delivering a low-cost, compact, and user-friendly glucose monitoring solution, the system enables individuals to engage in proactive health management, reducing reliance on centralized healthcare infrastructure. Furthermore, its modular architecture and ease of integration facilitate deployment in remote or home-based environments, supporting early diagnosis and intervention and improving long-term health outcomes.

4. Materials and methods

4.1. Sensor materials and structure

The core sensing element of the system is an EG-FET. The extended gate consists of a gold (Au) thin-film electrode, which is connected via a wire to the gate terminal of a MOSFET. This configuration physically separates the sensitive transistor channel from the electrochemical environment. The primary semiconductor device investigated was an N-channel depletion-mode MOSFET (LND150N3-G). The readout and control system was built around an Arduino Nano microcontroller. The analog front-end employed an LT1013DIP operational amplifier (Texas Instruments) and incorporated an ICL7660 voltage converter to generate the required negative supply voltage. All glucose-sensing experiments were performed using solutions prepared with PBS as the buffer matrix.

4.2. System design and readout circuit

The complete system, depicted in **Figure 2**, was designed as an integrated platform for sensor characterization and biosensing. It comprises four key functional modules.

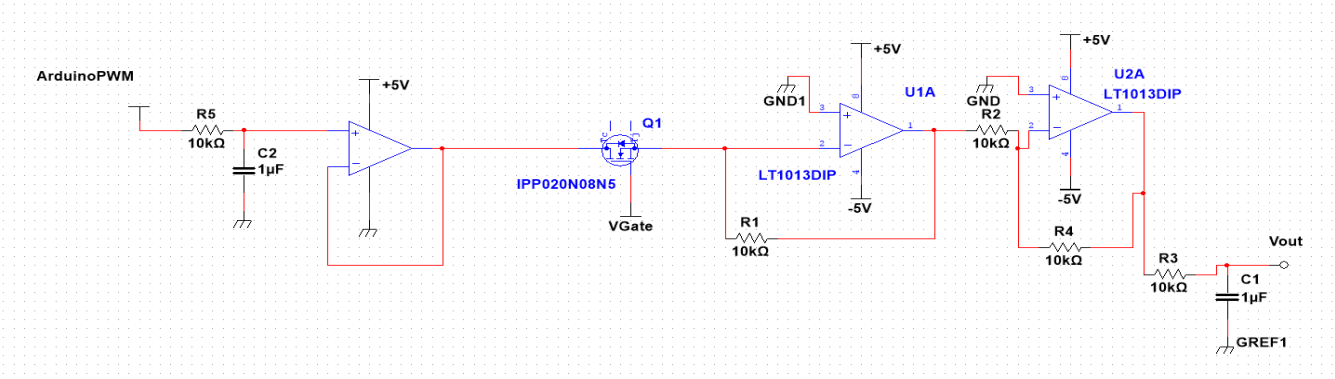

Figure 2. The system structure

4.2.1. Inverting op-amp circuit

Figure 3 illustrates the schematic of the inverting op-amp circuit. Resistor R1 is the feedback resistor, while R2 is the input resistor. According to the operating principle of the inverting op-amp circuit, the output voltage is given by: V_{out}-(Rin/Rf) *Vin.

In this system, the MOSFET replaces R2, so the source terminal of the MOSFET is maintained at 0 V due to the op-amp's virtual ground effect. Consequently, the drain current is determined by ID = -V_{out}/R1.

A 0–5 V PWM signal is applied to the MOSFET drain, and following the inverting amplifier configuration, the output voltage becomes negative. A second inverting op-amp circuit is introduced to convert this negative voltage into a positive signal compatible with the microcontroller's analog input, with R1 and R2 set to 10 kΩ.

This stage inverts the signal again, yielding a positive voltage suitable for signal processing.

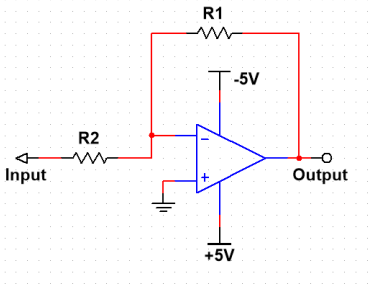

Figure 3. The structure of an inverting op-amp circuit

4.2.2. PWM low-pass filter circuit

Figure 4 shows the PWM low-pass filter circuit. The PWM signal is filtered through a first-order RC low-pass filter to attenuate high-frequency components. An operational amplifier is configured as a voltage buffer by connecting its inverting input and output terminals, leveraging the virtual short characteristic. This configuration stabilizes the filtered PWM output, ensuring consistent voltage levels for subsequent signal processing.

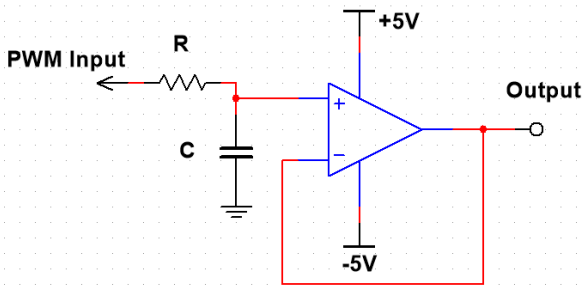

Figure 4. The structure of the filter

4.2.3. ICL7600 charge pump connection

Figure 5 illustrates the connection configuration of the ICL7660 module. This module's primary function is to convert a +5 V supply into −5 V, enabling the operational amplifier to receive a symmetrical ±5 V power supply required for proper bipolar operation. A 47 μF capacitor is included at the charge pump to accumulate charge and smooth the output voltage. Additionally, a second 47 μF capacitor is connected between the output and ground to suppress output ripple and enhance voltage stability.

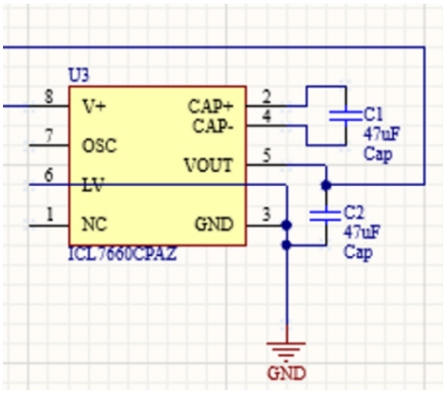

Figure 5. ICL7660 connection

4.2.4. Microcontroller interface and PCB

Arduino Nano was responsible for PWM signal generation and data acquisition. Analog pins A1 and A2 were used to monitor the drain-source voltage (VSD) and the output of the inverting op-amp circuit, respectively. A custom PCB was designed and fabricated to integrate all circuit components, with all traces routed on the bottom layer to facilitate manual soldering and debugging. The PCB design is shown in **Figure 6**.

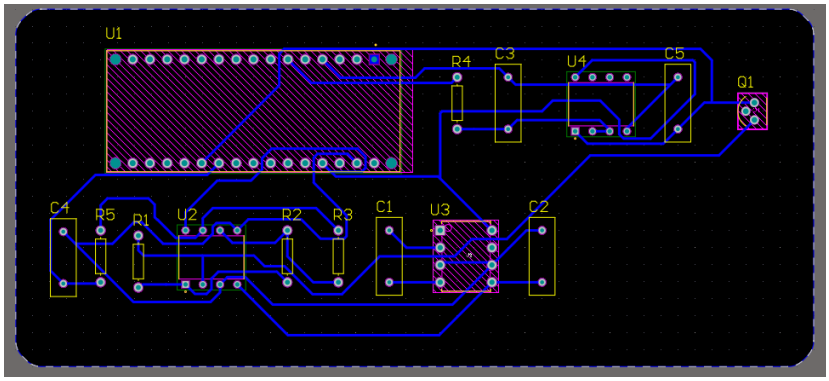

Figure 6. PCB design

4.3. Experimental procedure

4.3.1. Inverting op-amp circuit test

In **Figure 7**, an inverting op-amp circuit with a voltage gain of –1 was constructed, using an input resistor Rin and a feedback resistor Rf, both valued at 10 kΩ. A dual-polarity DC power supply provided ±5 V to power the op-amp. A sinusoidal input signal ranging from –10 mV to 10 mV was generated using a function generator, with test frequencies of 10 kHz and 100 Hz. The input and output voltages were monitored using an oscilloscope to observe and analyze the resulting waveform.

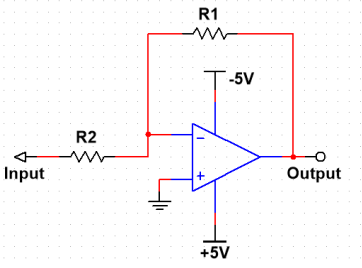

Figure 7. Inverting op-amp circuit

4.3.2. I–V characteristic test using a DC power supply

The source electrode of the MOSFET was connected to an inverting op-amp circuit, with a DC power supply used to apply a gate voltage ranging from –1 V to +5 V to emulate the potential of an extended-gate electrode. Another DC power supply was used to bias the drain terminal with a voltage sweep from 0 V to 5 V. A multimeter was connected to the output terminal V_{out}, and the feedback resistor Rf was set to 220 Ω. The measurement procedure was as follows: initially, V_{gate} was set to –1 V, and V_{drain} was incrementally increased from 0 V to 5 V while recording the corresponding V_{out}. Subsequently, V_{gate} was increased by 1 V, and the sweep of V_{drain} was repeated with new V_{out} measurements. This procedure was repeated across various V_{gate} levels to obtain the IV characteristics under different gate bias conditions.

4.3.3. The voltage converter test

According to the experimental manual [10], a voltage converter circuit was assembled. A 5 V DC voltage was supplied to the input terminal (V+), and the ICL7660 integrated circuit was tested using capacitors with values of 1 μF, 10 μF, and 47 μF, respectively. A digital multimeter was connected to the output terminal to monitor the output voltage of the ICL7660.

4.3.4. I–V characteristic test using Arduino Nano

Assembling the complete circuit, with a PWM signal generated by the Arduino Nano serving as the drain electrode voltage. A DC power supply was used to apply a voltage ranging from –1V to 5V to the gate electrode, simulating the potential of an extended gate electrode. While maintaining a constant gate voltage, the drain voltage was varied from 0 V to 5 V by adjusting the duty cycle of the Arduino's PWM output, and the processed data were recorded accordingly. Subsequently, the gate voltage was altered, and the previous measurement procedure was repeated to obtain the IV characteristics under different gate voltages as the reference data.

4.3.5. The gold film extended-gate test in glucose PBS solution

The gold film was connected to the gate electrode using an alligator clip. The experiment consisted of three control groups: Group 1 used pure PBS solution; Group 2 contained 1 fM (10^{-15} mol/L) glucose in PBS; and Group 3 contained 100 μM (10^{-4} mol/L) glucose in PBS. The gold film was first immersed in the pure PBS solution, ensuring that no part of the setup other than the gold film came into contact with the solution. At this stage, the PWM output was swept from 0 V to 5 V, and the corresponding processed data from the Arduino were recorded to generate the I–V characteristic curve.

Next, the solution was replaced with 1 fM glucose in PBS, and the measurement procedure was repeated. Finally, the same steps were performed using the 100 μM glucose solution. The I–V characteristics obtained from the three experimental groups were compared with reference data to establish the relationship between the drain current (ID) and glucose concentration gradients.

5. Results

5.1. The result of the I–V characteristic test without Arduino

The result is presented in **Figure 8**. The linear and saturation regions can be identified in the figure. A lower gate voltage leads to an earlier transition into the saturation region, consistent with the trends observed in the database [11].

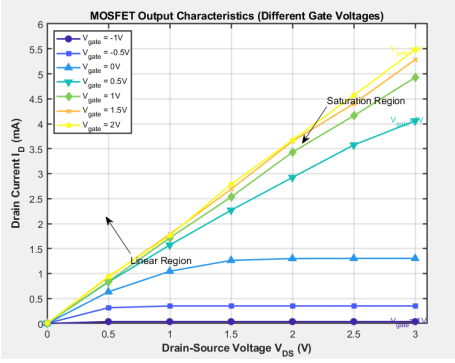

Figure 8. I–V characteristic test without Arduino

5.2. The result of the I–V characteristic test using Arduino Nano

The result is shown in **Figure 9**. The system, powered solely by Arduino, demonstrates reliable performance in completing the measurement tasks and yields a reasonable IV characteristic. This validates the system's suitability for further investigation of extended-gate electrodes.

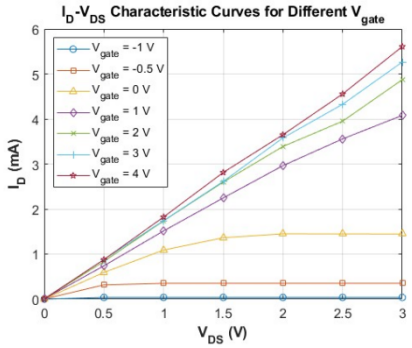

Figure 9. I–V characteristic test using Arduino Nano

5.3. The test of the gold film extended-gate test in glucose PBS solution

In this experimental test, in order to improve the measurement precision, the voltage step of the Arduino PWM output was adjusted to 0.01 V. According to the results, a fixed drain-source voltage of VDS = 0.8 V was selected within the linear operation region, and the relationship between the drain current (ID) and the glucose concentration in PBS solution was established. The result is shown in **Figure 10**.

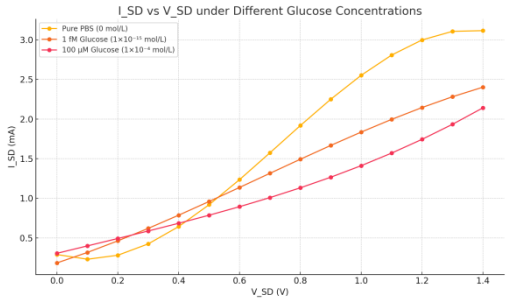

Figure 10. I–V characteristic under different glucose concentrations

6. Discussion

6.1. Interpretation of results

The project has achieved the following objectives. Firstly, an inverting op-amp circuit was constructed to measure the MOSFET IV characteristics accurately. By fixing the source electrode voltage at 0 V, the drain current is directly controlled by VDS, which leads to I–V curves. Secondly, a fully Arduino-powered EG-FET system was developed, demonstrating the advantages of miniaturization, low power consumption, and portability. Thirdly, A comparative analysis between the I–V characteristic test using the Arduino Nano and the I–V characteristics under different glucose concentrations revealed a linear relationship between the drain current (ISD) and the glucose concentration in PBS solution at a fixed drain-source voltage (VSD) within the linear region.

Although a reference electrode was not employed during the experimental process, using gold as a single electrode yielded meaningful results. Initially, a DC power supply was used to directly bias the gate

electrode, under which condition the resulting IV characteristics were accurate and could be used as a reference. Subsequently, the extended gold electrode was directly connected to the gate, and the potential generated on its surface was transferred to the MOSFET through mirror charge without any charge displacement, consistent with the first step.

In this project, it was observed that an increase in glucose solution concentration under identical VDS conditions decreases the drain current. When the extended gate of an EG-FET is implemented using a gold electrode, the surface potential of the Au-electrolyte interface becomes highly responsive to variations in analyte concentration. Increased glucose concentration increases interfacial potential due to enhanced ionic activity or metabolite adsorption. This elevation in interfacial potential effectively reduces the gate-to-source voltage, lowering the drain current ID. The observed logarithmic relationship between ID and glucose concentration is consistent with the Au surface's Nernst-like behavior and supports the EG-FET biosensors' electrochemical gating mechanism [12].

6.2. Stability and error analysis

In this project, multiple non-DC voltage signals are involved. The first is the ICL7660 charge pump, which achieves voltage inversion by utilizing the charge stored in a capacitor and outputting it in an inverted manner. However, due to its intrinsic limitation of supplying only a limited output current, it is highly susceptible to load variations, resulting in output levels that often fail to reach the nominal 5 V and are superimposed with ripple. This may adversely affect the performance of operational amplifiers powered by this source, especially under asymmetric supply conditions, potentially leading to restricted output voltage range and increased distortion, which is particularly detrimental in high-precision applications [13]. Secondly, the Arduino-generated PWM signal is considered. Under typical conditions, Arduino PWM outputs' frequency and duty cycle are relatively stable. However, slight jitters may still occur due to the constraints of software timing and clock accuracy. Although an RC low-pass filter and op-amp buffer are employed to process the signal, potential issues remain.

6.3. Limitations

Firstly, an inverting op-amp circuit was employed to test the IV characteristics. Due to the limitations imposed by the experimental setup's safe current range, the op-amp's gain could not be set too high, which restricted the system's ability to detect minor voltage variations. Secondly, when testing glucose solutions in PBS using a gold extended gate, the bare gold electrode was prone to contamination, oxidation, and mechanical instability during immersion. Moreover, the experimental environment was suboptimal, which increased the risk of cross-contamination in glucose solution samples that were repeatedly measured.

6.4. Project impact and contribution

This project presents a fully functional, low-cost EG-FET-based glucose sensing platform that integrates key subsystems, including a chemical-to-electrical transduction layer, analog signal conditioning circuitry, and a digital data acquisition and communication interface. The system demonstrates how a depletion-mode MOSFET, combined with an extended gate electrode, can construct a reliable and highly responsive biosensor without reliance on enzymatic components. An inverting operational amplifier circuit simplifies the signal pathway while maintaining sufficient accuracy to discriminate glucose concentration levels.

7. Conclusion

The project designed and developed an EG-FET glucose sensor fully powered by an Arduino Nano. Comprehensive investigations explored the characteristic behaviors of operational amplifiers and MOSFETS. A gold extended gate was employed to interface biochemical signals with electrical signals, which were subsequently processed and visualized using the Arduino platform. A linear dependence of the drain current on glucose concentration was observed at a constant drain voltage within the linear region of operation. The observed signal behavior conformed to the theoretical expectations of field-effect modulation, thereby validating the feasibility of the EG-FET architecture for non-enzymatic sensing applications.

Disclosure statement

The author declares no conflict of interest.

References

[1] Turner APF, 2013, Biosensors: Sense and Sensibility. Chemical Society Reviews, 42(8): 3184–3196.

[2] Poghossian A, Schöning MJ, 2018, EGFET-Based Sensors for Bioanalytical Applications: A Review. Sensors, 18(10): 3575.

[3] Kim Y, Hong J, Kim DH, 2023, Extended-Gate Field-Effect Transistor Chemo- and Biosensors: State of the Art and Perspectives. Sensors and Actuators Reports, 5: 100108.

[4] Li X, et al., 2017, Porous Gold Films—A Short Review on Recent Progress. Micromachines, 8(6): 159.

[5] Sharma S, Saikia AS, Kalita PK, 2021, Generations of Glucose Biosensors. Encyclopedia, 1(2): 387–402.

[6] Bergveld P, 1970, Development of an Ion-Sensitive Solid-State Device for Neurophysiological Measurements. IEEE Transactions on Biomedical Engineering, BME-17(1): 70–71.

[7] Kaur R, Gupta P, Singh J, 2018, A Review on Power MOSFET Device Structures. International Journal for Research in Applied Science and Engineering Technology, 6(4): 1234–1240.

[8] Lee J, et al., 2021, A Review of BioFET's Basic Principles and Materials for Biomedical Applications. Micromachines, 12(4): 319.

[9] Ramdeo JT, Srivastava VM, 2023, Prototype Design and Analysis of 4-Stage Operational-Amplifier using Double-Gate MOSFET. International Journal of Engineering Trends and Technology, 71(7): 175–188.

[10] Intersil Corporation, 2005, ICL7660 CMOS Voltage Converter, Datasheet, viewed April 15, 2025, https://www.alldatasheet.com/datasheet-pdf/pdf/541063/INTERSIL/ICL7660.html

[11] Microchip Technology Inc., n.d., LND150 – N-Channel Depletion-Mode MOSFET, Microchip Technology, viewed April 15, 2025, https://www.microchip.com/en-us/product/LND150

[12] Tarasov A, Wipf W, Stoop M, et al., 2012, Signal-to-Noise Ratio in Dual-Gated Silicon Nanoribbon Field-Effect Sensors. ACS Nano, 6(10): 9291–9298.

[13] Jung W, 2005, Op Amp Applications Handbook, Newnes, Burlington, MA.