

Advanced FET-Based Biosensors—A Detailed Review



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Abstract This review paper has done a detailed performance analysis for advanced field-effect transistor (FET) based biosensors which shall be distinguished by their outstanding features, such as mass-production capability, sensitivity, ultra-sensitivity detection, and low-cost manufacturing, within a range of advanced biosensing technologies. In order to encourage the understanding of FET-based biosensing technology and its sensing analyte, major FET-based biosensing devices are presented in this review: Dielectric modulated FET, impact ionization MOSFET, double gate -dielectric modulated tunnel FET, junction less electrolyte insulator-semiconductor FET and nanowire FET, etc. This work is also designed to provide a state-of-the-art analysis of biosensors, based on an advanced field-gate field-effect transistor in the area of bioanalytical applications. Besides, a connection will be made between the various FET structures, with particular attention paid to materials and technologies.

Keywords Field-effect transistor · MOSFET · In-vitro diagnosis (IVD) · Biosensors

1 Introduction

Micro/Nanotechnologies are emerged with ultra-sensitive biosensors. Due to early detection potential, the advent of specificity medicine, genetic testing, and gene sequencing attention is drawn. Academic biosensing explores practical tools for in-vitro diagnosis (IVD) as distinct as you are, sensing methods are developed, and these advances translate. Simultaneously, analysis strips and instruments are being developed in a ground-breaking transition from electrochemical / optical to nano-electronic technologies. It is also worth looking at electronic-based biosensing instruments, one

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of the leading medical diagnostic sensing technologies. FET-based biosensors have been proposed and have become an emerging field due to solid-state technologies' rapid growth. Biomolecules are used to carry electrostatic charges where the bioactivities are needed for electrostatic potential adjustments. A FET-based biosensor would be a healthy choice for ultra-sensitivity and rapid reaction conditions.

2 Biosensors Using FETs-Related Works

The study of embedded field-effect transistor underlap channel submitted by Jee-Yeon Kim for biosensor application [1]. For both aqueous and dry electrical label-free biosensors, a low-powered FET channel is suggested, and the voltage/current properties are compared and evaluated in each collection. Avian influenza (AI) antigen-antibody binding is used to research the underlap framework's efficacy as a biosensor for both settings. The underlap-FET proved to be label-free; biomolecules were electrically defined, and the two conditions contrasted in both aqueous and dry conditions were quantitatively compared. When the anti-AI is spring to the underlap channel region enclosed by SBP-AI, the decrease in drain current is calculated using the negative charges of anti-AI molecules. This effect has been seen for both conditions, and the drain current change was greater in dry conditions than in aquatic environments.

"Study the performance of Dielectric modulated FET" was presented by Kannan, member IEEE [2]. In this paper, using computer-aided simulation technology as a label-free biosensor, analyses the outcome of a dielectric modulated I-MOS transistor (DIMOS). By inserting a nanogap into the I-MOS device by mixing chromium and gold gate structure, DIMOS provides the extremely sensitive with potential biomedical and biomolecule sensing applications. Impact ionization dielectric modulated based biosensor FET has been used to study the nanogap immobilization biomolecular detection and instrument's sensitivity. In the simulation, the nanogap's dielectric constant is modeled for immobilization, and even the dielectrical constant of biomolecules would be separate from the air. A value of $K = 2$ and 12 is used to define a category of biomolecules with low- and high-dielectric constants. The notion of a DIMOSS has been suggested and studied. With a biosensor. The recommended structure shows high sensitivity. Dielectric modulation is the dominant effect of biomolecule detection. It retains dielectric modulation compared to conventional dielectric modulated FETs, where the effect on interface sensitivity is essential for biomolecule charges.

Ultra-sensitive sensing performs by using vertical strained impact ionization MOSFET presented by Ismael Saad [3]. Interestingly, the vertical strained impact ionization MOS suffers from an impressive high-supply voltage (VDS) and hysteresis. Hence, the idea of strained SiGe vertical I-MOS is implemented to reduce supply voltage. As the strained SiGe layer is inserted into the framework, both the supply and threshold voltage decreased significantly. After all, the system continues to experience low-breakdown voltages caused by parasitic bipolar transistors (PBT)

that affect the device's reliability. Developing novel system designs to replace traditional I-MOS is a potential solution to address these limits. For potential applications biosensing, the performance of three attractive candidates was investigated and defined: vertical strained impact ionisation MOS dual channel (DC-VESIMOS), vertical straight-in ionisation MOS single-channel (SC-VESIMOS), and vertical stress ionisation MOS Dielectric Pocket (VESIMOS-DP) integration. Comparison of transfer characteristics (IDS-VGS) with SCVESIMOS for a candidate with the best possible biosensor for characterizing the respective unit dual channel vertical stressed ionisation MOS's performance and vertical stressed impact ionisation MOS-Channel length dual pocket devices $L_g = 50$ nm. The voltage sub-risk of VESIMOSDPP increases $I_{off} = 10^{-16}$ A/ μm leaking currents and ON current of $I_{on} = 10^{-4}$ A/ μm single-channel-VESIMOS. Therefore, a low-leakage current gives low subthreshold advantages. The system offers high stability and low-energy consumption. The DC-VESIM/OS method is the best candidate for the potential lowest biosensor sensitivity software.

Saiyan Kanungo [4] presents the efficiency comparison of the full and short gate TFET with different dielectric values. In this paper, they studied the efficiency and fundamental operational physics of both FG-DMTFET and SG-DMTFET dependent biosensors. The schemes of DG Dielectric Modulated Tunnel FET biosensors are shown in Fig. 1 a, b.

Silicon–Germanium acts as the basic material for the higher tunneling current value in both device structures with a germanium composition of 0.5. The gate length of the full gate-dielectric modulated tunnel FETs is known to be 42 nm, while the design of the short gate-dielectric modulated tunnel FET is 20 nm. Each of the drain and source regions is 20 nm long, and the channel thickness is considered to be 10 nm. The source ($p+$), channel (P), and drain ($N+$) regions have uniform doping concentrations are range between 10^{19} cm^{-3} and 10^{12} cm^{-3} . The bias dependency of tunneling junction electrostatics from such biosensors should be undertaken with a comparative analysis better to understand dielectric modulated tunnel FET biosensor's relative performance. The implementation of the shorter gate tunnel FET with dielectric modulated and full gate-dielectric modulated tunnel FET biosensing components has been extensively studied. Compared to the traditional FG architecture, the integration of the short gate architecture into the dielectric modulated tunnel FET framework provides a substantial increase in sensitivity without substantial manufacturing. The short gate biosensor can work within a precise range of biases. The short gate-dielectric modulated tunnel FET architecture will generate drain current in the simulation context similar to the streptavidin–biotin binding system (i.e. $k = 2.1$ and $\text{son} = 0$) and its found that approximately seven times improved sensitivity of short gate TEFT compared to Full gate DM Tunnel FET.

“Study of SiGe and pocket-doped sensing activity effects” was presented by Partha Sarathi Gupta [5]. Compared to its dielectrically modulated FET counterpart, biosensors based on dielectrically modulated tunnel FET (DMTFET) exhibit greater sensitivity, but it has low subthreshold characteristics. The impact of the use of the source silicon–germanium (SiGe) and the channel $n+$ -pocket-doped is

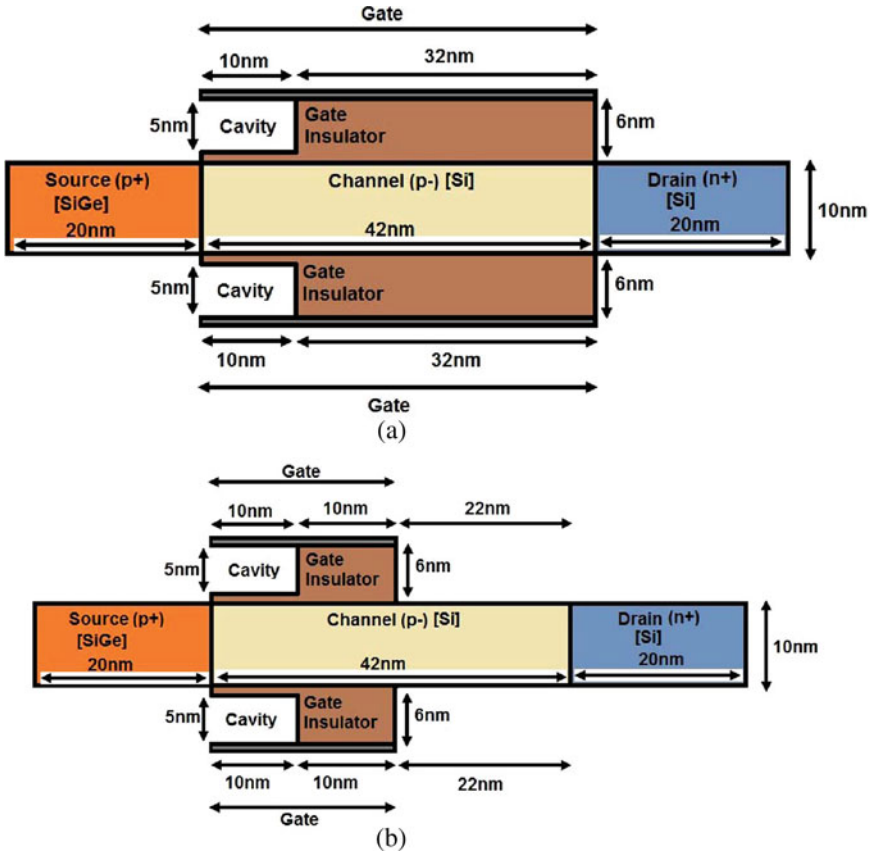
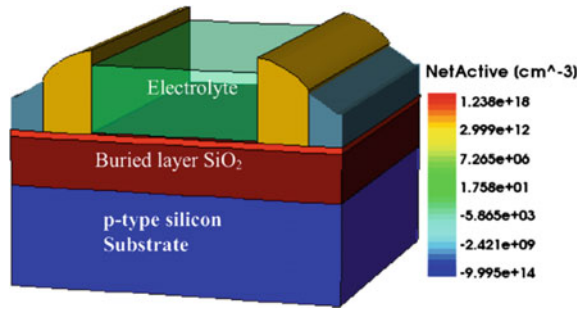


Fig. 1 a DGDM TFET biosensors with full Gate and b short gate DGDM tunnel FET biosensors

discussed in this context. For biomolecule conjugation, variation in doping concentration in the $n+$ -pocket region, the source region, and germanium composition's fundamental mechanics are discussed. The decrease in source side bandgap and its effect on band-to-band tunneling has been studied in a reduced tunneling distance. The sensing efficiency of dielectric modulated tunnel FETs was measured subsequently. Research shows that SiGe source DMTFET is much higher than $n+$ pockets to achieve higher current subthreshold levels, while retaining the necessary sensitivity. This sensitivity-current optimization was investigated, and the required bias function was indicated for different gate voltages and drain variations. Each source and drain area should be 50 nm. The insulator channel and gate thicknesses are 20 and 10 nm, respectively. Suppose to optimize the electrical reaction of the conjugation, SiO₂ and aluminum (work function = 4.1 eV) were gated. Doping concentrations are consistent, respectively, of $1 - 10^{20}$, $1 - 10^{16}$, and $5 - 10^{18} \text{ cm}^{-3}$ in the $p+$ source, p -channel and $n+$ -drain regions. The incorporation of SiGe source into the dielectric modulated FET device provides a major advantage over the pocket-doped

Fig. 2 Simulated n -type junction less electrolyte insulator-semiconductor FET



architecture for sensitivity optimization. The use of SiGe source offers more than one order of improvement for a wide variety of biomolecule samples without substantial trade-off sensitivity, while the $n +$ pocket dielectric module tunnel FET has seen a consequence of more than 5% sensitivity degradation to achieve a proportional current improvement.

“Junction less electrolyte insulator-semiconductor FET details” was presented by Ajay [6]. The TCAD simulation of n -type junction less electrolyte insulator-semiconductor FET is shown in Fig. 2. In typical MOSFET, the gate material is directly connected to the gate oxide (SiO_2 insulator). In ion-sensitive FET, the measuring electrode, also called the reference electrode (V_{ref}), is placed around electrolytes into the dielectric layer (oxide layer). The ions present in the electrolyte and its charged molecules influence the gate terminal’s electrostatic potential, so they can adjust the threshold voltage and drain current. An essential element of the ion-sensitive FET is the electrode. It provides the test electrolyte with a stable electrical contact and allocates the sensing liquid’s electrical potential or the sensor’s operating point depending on the FET. They may also exchange information with electrodes and electrolytes, and the drainage current of the system depends on the electrolyte arrangement. Thus, by changing the drain current is a change in the gate’s surface potentials at the gate isolator and electrolyte interface.

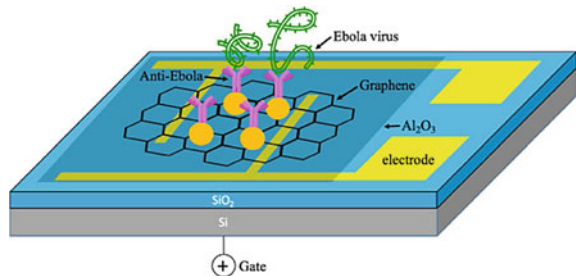
The conductance of the junction less silicon-on-insulator electrolyte semiconductor FET channel depends not only on the device operating, but also, on electrolyte status and solution’s ion concentration. Different pH values have been demonstrated to affect the conductance of the junction less FET. As the electrolyte’s hydrogen ion concentration increases, the pH value of the electrolyte is also modified. This also affects the junction transfers less silicone electrolyte semiconductor FET. The Silicon-on—insulator junction less (JL) electrolyte semiconductor field effect transistor (EISFET), a well-known 3-D commercial TCAD semiconductor, was investigated as a simulation method. The area of the electrolyte consists of altering the properties of the pure semiconductor material as the electrolyte is the same as the pure semi conductive material. A solution determined by the necessary carrier load-densities of the intrinsic semiconductor material with efficient phosphate-buffered Salin (PBS) and ionic concentrations was implemented. Michael Kroon [7] presents fast detections of the Ebola antigen with FET. For real-time Ebola virus antigen

detection, a reduced field-effect transistor device based on graphene oxide. This method utilizes the graphene-based smart semiconductor features and instantaneously produces highly sensitive and reliable Ebola glycoprotein detection. The schematic diagram of the graphene oxide-based FET biosensor is shown in Fig. 3.

The graphene oxide layer was dumped on the instrument to link the source and the drain regions electrode. A thin layer Al_2O_3 is coated for surface passivation on the graphene oxide layer. The transistor measurement was performed to describe the essence of the semiconductor and the start-up current ratio. The electrodes of drain and source region were given a constant 0.01 V of V_{ds} with V_{gs} between -40.0 and $+40.0$ V. The dynamic sensor response to EGP was investigated with constant V_{ds} of 0.01 V across drainage and supply electrodes in order to examine the sensor output and V_g set at zero. Higher V_{ds} leads to a louder reaction and even damage to the rGO sheet. EGP was taken in the form of an industrial supplier's purified protein (IBT Bio Services) and interrupted into 0.01PBS /human/human plasma. As PBS/serum/plasma dilution ensures that the surface load transmitted by EGP is not screened the Debye duration increases. The respective length of Debye is around 7.4 nm 35 for PBS, while the gold NPs are less than 5 nm and the provider reports that Ebola and EGP antibodies are about 3 and 10 nm.

Yu-cheng-syu [8] is addressed with a field-effect transistor analysis with a clinical use. A biosensor is a device used by the International Union of Pure and Applied Chemistry (IUPAC) to detect chemical compounds by electrical, optical signals or thermal using biochemical reactions intercede by immune systems, isolated enzymes, tissues, and entire cells. In other words, to track the dynamics and interactions of the immune system, isolated enzymes, organelles, tissues, and whole cells, a biosensor is an analytical tool. DNA hybridization transfers the results of tracking into electrical signals. The FET-based biosensors were targeted to be the right candidate for POCT testing of the next generation. FET-based biosensors are planned to be an acceptable candidate for POCT (Point-of-Care) research of future generations and an IVD subdivision. It is characterized as a medical diagnosis at the point where patient care is required. For example, a glucometer is used as a central laboratory in the home of the patient instead of in health centers. It should be simple enough for less qualified people to work for POCT or not even trained. In addition, it greatly reduces the time taken for test results, providing doctors with immediate diagnostic information. Diamond field-effect transistor (FET) a solid surface sensing feature

Fig. 3 The reduced graphene oxide-based FET biosensor schematic diagram



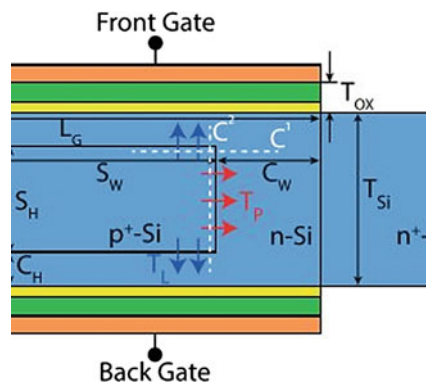
using ribonucleic acid aptamers (RNA). The results show a possible change at the 91.6 mV gate, whereby 8 μ A shifts occur in the negative path when HIV-1 Tat proteins are present in the source-drain connected with RNAA protein. In addition, it was first demonstrated that aptamer-FET has been able to rely on the actual HIV-1 sample of Tat proteins, which demonstrated a good approach to the development, through diamond bio-interfaces, of relevant clinical biosensor applications.

“New dual pocket vertical heterostructure TFET performance” was presented by Amit Bhattacharyya [9]. The evolution of FET-based biosensors could be successfully realized by high-throughput biomolecule recognition technology. Medical diagnosis is helpful, but implementations often have certain drawbacks. Due to the effective detection, area saturation, size, and molecular concentration issues would restrict the response limit among various types of biomarkers. The cavity with a low-binding probability, surface functionality was included in the proposed architecture. So on, the collection of bio analytes inside the cavity is not smooth and complex. A more practical strategy would be this one. Also, the filled cavity is not allocated to the steric hindrance problem. Thus, different non-uniform phase patterns are considered inside the cavity.

The following International Technologies, the incorporation of regular/irregular biomolecules arrangements was proposed as a semiconductor roadmap. The proposed dual-gate TFET structure would be used to analyze the level of the well-setup unit easily. Channel length (L_{ch}) = 42 nm the source and drain regions are fixed as 20 nm, and the cavity length is fixed as 15 nm are the simulation parameters. This paper shows a new FET-based biosensor with a double-pocket-dielectric modulated heterostructure tunnel involving lateral and vertical tunneling events. In comparison to other recognized TFET structures, the efficiency of double pocket-hetero-TFET (HTFET) was assessed.

Sandeep Kumar [10] is provided for the dual-gate operations conducted using an extended source double gate (ESDG)-FET, and the schematics are shown in Fig. 4. Also, the authors analyses the comparative studies for single and dual-metal gate DMTFET. Besides, a double gate-dielectric modulated tunnel FET-based biosensor

Fig. 4 Schematic view of ESDG-DMTFET



was recorded in the source region over which a single-sided *n*+ -pocket DG-DMTFET-based biosensor was applied. In order to achieve improved sensitivity, the cavity was extended. A staggered DMTFET heterojunctions biosensor was suggested to give the threshold voltage sensitivity (V_{th}) of 450 mV. The source is enlarged further into n-channel so that both sides overlap the source. The role of gate metal work (ϕ_m) and length of the gate (L_G) 3.8 eV and 90 nm, respectively, are recognized. Field plate is used to increase tunneling and reduce channel resistance at source-channel junction. The (L_{ch}) value is chosen so that the extended source double gate -DMTFET configuration enables line and point tunneling at the source-channel junction. It boosts current sensitivity, and biosensor threshold voltage, drain current.

This presents the ESDG-DMTFET biosensor in the channel region with an extended source. Biomolecule immobilization cavities are formed in drain side gate-oxide. Changes in k of biomolecules allow for the full change of the I_{on} and V_{th} system. From the source-channel interface with an increase in k , the electrical field increases, this reduces the tunneling width and increases the I_{on} and I_{ON}/I_{OFF} sensitivity of the above device structure. However, from the other performance, V_{th} reduce and I_{ON} rises due to the increase in k .

Chattopadhyay [11] presents a wide range of oxide stack detection for junction less FET (Fig. 5). The performance of a dielectrically modulated low and high- k oxide stack junction dual-metal double-gate was investigated with low HK-S JL-MOSFET and DM-DG-LK biosensor system for the effective recognition of various protein molecules in a dry condition, in terms of absolute and relative change, called V_{th} -responsiveness and threshold voltage (V_{th}) V_{th} -sensitivity.

For JL-MOSFET-based biosensing applications, the sensor with a long channel length is desired in the present situation, as it is preferable to connect sufficient numbers of biomolecules to the sensing device. The nanoscale regimen's surface is relatively thin. Unless otherwise reported, all our research considered a JL-MOSFET 1 μm channel length. However, shorter channel lengths are necessary for better system efficiency, especially for digital applications. Two different channel length

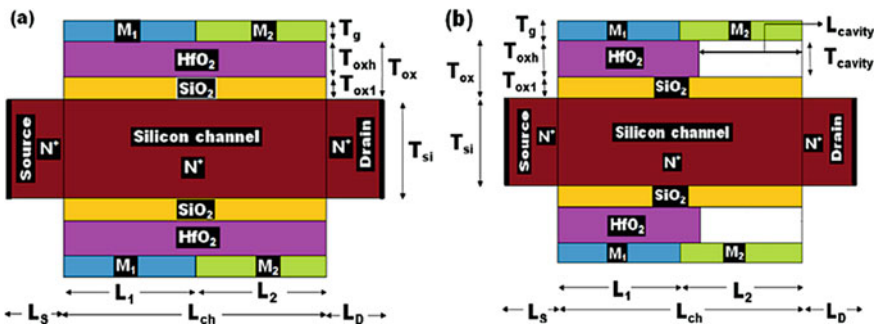


Fig. 5 a 2-D DM-DG-LK/HK-Stack JL-MOSFET cross-sectional view, b DMDG-Low-K/High-K-Stack Junction less-MOSFET schematic with Nanogap cavity

values, i.e., 1 μm and 50 nm, were performed in the dielectric modulated-double gate-low-K/high-K-stack junction less-MOSFET as a sensor system for the detection of protein molecules.

Louisa sellami introduces fluid-biosensor efficiency [12]. This paper consists of a gap-sensor with a bridge contains three regular PMOS devices attached as resistors and condensers, three bridge brackets. The fourth arm is a same PMOS transistor, and it distinguishes between a channel and a high polycap where a fluid like creatinine or gas is located. The circuit method uses spice simulations and spice extractions. In this, a sensor-sensitive is given to a dielectric fluid constant over a constant dielectric range of 11–1. This is done with a normal VLSI analog processing, which partially removes the thin dielectric gate of one of the four bridge bin transistors for which additional oxide grafting is applied. The V_{set} value of the bridge, representing the relatively dielectric constant of the fluid in silicon dioxide, can be compared by adjusting the V_{set} diagonal transistor gate.

Pranav Amborkar [13] describes the growth of the nano wire technology and in Fig. 6 the scheme is shown. High-sensitivity biomedical sensors may make it possible to identify diseases in their early stage, dramatically raising the likelihood of diagnosis and action that could potentially save lives. For example, five years' survival rate is higher than 90% when breast cancer is diagnosed and treated with current therapies at an early stage (local condition), but drops to about 20% when the late stage (distant disease) happens. Early detection of cancer also requires highly sensitive biosensors. Although much research has been carried out to boost the sensitivity of biomedical sensors, recent development in the field of nanotechnology will deliver the most promising biomedical sensor solutions. Nanotechnology encompasses large fields

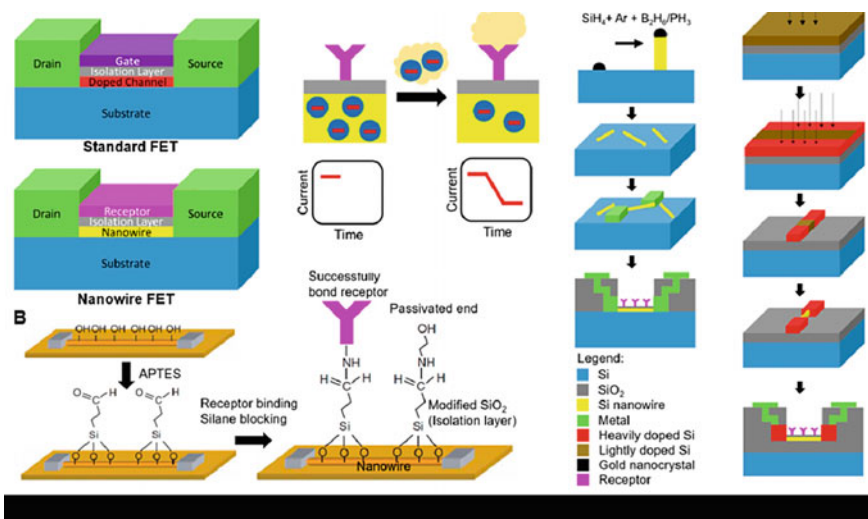


Fig. 6 Nanowire technology for bio medical sensors

of study in science and engineering in order to investigate materials and structures below about 100 nm.

Nanowire sensors have shown great dedication to being available sensors—the biological and medical recognition network. The devices offered to offer many advantages, such as high-sensitivity electrical signal transduction in real-time and label-free detection feasibility. While it is remarkably sensitive to its analytical signal capabilities, it is still too low for other methods, often used in vivo settings, to be polluted by high-background noise. This greater sensitivity and more fundamental problem of development can be solved by improved receptor binding methods. Moreover, the higher yield of recent up-to-date processing techniques allows lower prices of marketed products. However, nanowire sensor efficiency is based on their advances in simplicity, sensitivity, precision, and durability compared to existing standards in the goldfield, such as enzyme-linked immunoassay (ELISA) and polymerase chain reaction (PCR).

“The Chemically functionalized graphene FET” was presented by Clare watts [14]. Graphene FET (GFET) was covalently functionalized with 1-pyrene butyric acid and paired with anti-CD63 antibodies for N-hydroxy succinimide ester exosome-free detection. The microfluidic tube solution revealed a portion of the graphene film. In addition to the original Dirac point in the background voltage (V_g) curve, the electrical properties of the diagram were further reduced. In the pipe, a minimum of one V_g was less than the initial Dirac stage when the phosphate saline (PBS) was present and, over time, changed by injecting exosomes into the tube. This minimum change from PBS was saturated after 30 min, and many exosome levels were observed. The highest concentration of exosome in combination with the anti-CD63 antibody means that functionally functionalized GFET can directly detect exosomes low to a minimum of 0.1 $\mu\text{g/mL}$ and susceptible to concentration, was low in combination with the isotype regulation (Fig. 7).

Fig. 7 Microfluidic integration biosensor

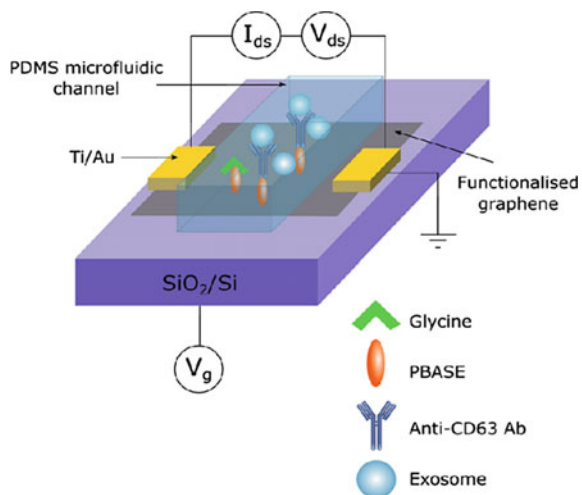


Table 1 Performance analysis of different FET based

Reference number	Different types of FET	Channel length (L_{ch}) (nm)	source -Drain voltage (V_{ds}) (V)	Source-gate voltage (V_{gs}) (V)
[2]	DIMOS	200	–	–
[3]	VESIMOS	50	1.75	1.75
[4]	DM-TFET(FG&SG)	–	0.4	1.2
[5]	DMTFET	100	0.5	6
[6]	JL-EISFET	5	0.5	1.3
[9]	HTFET	42	0.8	1.3
[10]	ES-DG-TFET	15–40	1	1.5

3 Major Findings

It has been inferred from the presented literature survey that, HTFET, ES-DG-FET privilege optimum performance for nano scale biosensor. The relationship between VDS, VGS has been plotted in Table 1.

4 Conclusion and Future Scope

The broad advancement of nanotechnology has rendered many possibilities feasible in the area of nano biosensors. In this paper, several FET structures with respective design specifications and performance metrics have been discussed. The biosensor should discuss settling time, sensitivity, and selectivity to ensure optimum operation. However, in terms of sensors' development, many aspects should be carefully examined, including biomolecular binding energy or the screening issue. There should be a compromise between the sensitivity of the sensor and stability of receptor-target molecules. In future, nano wire, extended gate FET, nanodot sensor, flexure FET sensor are may be used to design the biosensor for reduce the channel length and low power application.

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