

Analysis of the thermal stability and sensitivity of graded-channel split-gate nanotube Junctionless FETs for dry-state biosensing

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The present study investigates the sensitivity behaviour of a Junctionless graded-channel split-gate nanotube (NT) field-effect-transistor (FET) based biosensors for the detection of label-free biomolecules over a temperature range of 250–500 K over a temperature range of 250–500 K. While optimal biosensing sensitivity is observed within the physiological window (<320 K), the temperature was extended to 500 K to demonstrate the thermal reliability of the sensor platform. The proposed Bio-NT-JLFET employs a silicon nanotube FET operating in accumulation mode with graded channel doping. For smooth sensing of biomolecules and enhanced stability, a gate underlap double-sided cavity has been created in the Bio-NT-JLFET. Surface-charge densities for biomolecules were set to realistic ranges ($10^{-7} - 10^{-8} \text{ Ccm}^{-2}$) with corresponding dielectric constants, representing dry/low-ionic operation. The drain current and subthreshold swing sensitivities are higher for double-strand (ds) DNA, while threshold voltage sensitivity is higher for Al-ab biomolecule relative to the air reference. All simulations were carried out using a calibrated TCAD framework validated against a reference Si-NT-FET structure on ATLAS TCAD device simulator. The results establish the Bio-NT-JLFET as a robust candidate for ex-situ dry-state detection.

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1. Introduction

Deoxyribonucleic acid (DNA) molecular analysis allows the direct diagnosis of numerous contagious diseases caused by various strains of type-A influenza viruses and genetic mutations [1, 2]. Besides enhancing the chances of recovery, it lessens the associated complications and reduces the risk of virus transmission. DNA involves two molecules of single-stranded DNA (ss-DNA) enfolded together in a helical fiber called the double-stranded DNA (ds-DNA). The ss-DNA consists of a linear structure with only one DNA strand, while the ds-DNA forms a helical structure with two DNA strands connected like hydrogen bonds [3]. Nowadays, as most biomolecules possess electrostatic charges and the related bioactivities include variations in electric potential, the field-effect transistor-based biosensor is a popular choice to observe and predict dynamics and exchanges of biological activities, e.g., DNA hybridization and cell activities [3]. It transfers the observed results into a measurable electric signal (i.e., current or voltage). The biosensors exploiting nanogaps-etched FETs [4][5] are increasingly favoured for label-free detection of neutral or charged biomolecules. These nanogaps, embedded within the oxide layers, capture the biomolecules and influence

the threshold voltage (V_{th}) and on-current (I_{on}) of the device on the basis of the dielectric constant and electrical properties of biomolecules, enabling the precise detection of specific biomolecule types. However, it has been found that the proximity of the gate electrode to the nanogap can negotiate the structural stability of the device. Barik et al. have explored an underlapped FET to resolve structural instability problems [6], where the sensing area is created by etching a substantial percentage of the layers of metal-gate and oxide. In this way, the retained thin gate oxide layers act as a binding site in the nanogap and underlap regions of the structure under consideration [7]. A thin gate oxide is retained in the nanogap/underlap region to strengthen the structure [7]. To further improve transport efficiency and to diminish the hot carrier effect, gate metal work-function engineering (GMWE) technique has been utilized in nano-wire FETs by Y. Pratap et al. [8] and A. Das et al. [9].

In the ever-evolving landscape of electronics, Junctionless field-effect transistors (JL-FETs) play a pivotal role. Unlike conventional MOSFETs, JL-FETs do not have metallurgical junctions, making them simpler to process and less costly to manufacture [10]. They utilize a gated semiconductor film to control resistance and current flow [11]. The cylindrical structure of Gate-All-Around

nano wire and Double Gate-All-Around nano tube (NT) provides superior control over the channel compared with cuboidal 3D structures [12-14]. High packing density, superior control over the channel, and steep subthreshold slope are some of the outstanding aspects that attract the attention of researchers towards nano-wire (NW) structures. Fahad et al. [15] have put forward that the nanotube (NT) with two cylindrical gates is superior as they have high drive-current, high *on*-current, and *off*-current (I_{on}/I_{off}) ratio, and low inverter delay time compared to NW structure. A. Kumar et al. [16] had shown that for tube thickness less than 4 nm, the mobility of carriers was enhanced in NT devices and reflected a significant improvement in drive current. In another work, it has been demonstrated that the transconductance and packing density can be improved by employing the split-gate technique [17], whereas *on*-current (I_{on}) and cut-off frequency can be further enhanced by using high-*K* gate material and graded junction technique, respectively [18-19].

In the present work, an enhanced temperature-sensitive biosensor for avian influenza antibody (AI-ab) and DNA biomolecules has been proposed by

incorporating structural engineering, such as split gate (SG) and graded channel (GC), in the nanotube Junctionless double-gate FET (Bio-NT-JLFET). The graded channel is employed to suppress the *off*-current, thereby improving the sensitivity of the sensor [20], [21]. The considered temperature range has been taken between 250 K and 500 K. The performance assessment of the Bio-NT-JLFET device has been executed in terms of variation in sensitivity against threshold voltage, drain current, subthreshold swing, *ss*- and *ds*-DNA, and other important characteristics with respect to temperature variation on device simulation software. The ensuing sections of the article include the sections devoted to discussing the proposed device structure and method, followed by analysis of simulation results, and conclusive notes.

2. Device structure and methods

The three-dimensional (3D) schematic diagram of the simulated graded-channel split-gate nanotube Junctionless FET-based biosensor (Bio-NT-JLFET) is shown in Fig. 1.

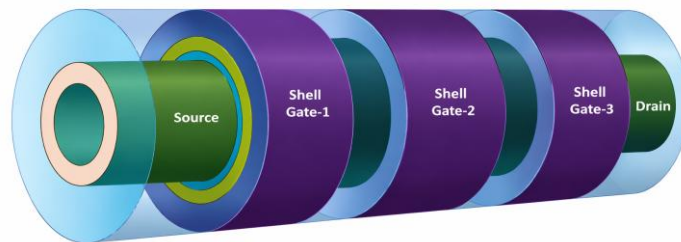


Fig. 1. Three-dimensional view of the Bio-NT-JLFET sensor

The two-dimensional (2D) horizontal cut-plane view of the cylindrical device incorporates the channel, inner-gate oxide, and outer-gate oxide, source, drain, and metal-gate contacts as displayed in Fig. 2 (a). The channel region is an ultrathin silicon nanotube (NT), which integrates an inner-gate placed inside and outer shell-gates casing the same from outside. A thin layer of SiO₂ is grown around

the entire channel, above which three separate layers of HfO₂ (high-*K* dielectric) are deposited side by side to split the gate and constitute the split gate (SG) structure. In this way, two cavities get opened near the source and drain sides, respectively, for introducing the biomolecules. The graphical representation of key fabrication steps is shown in Fig. 2(b).

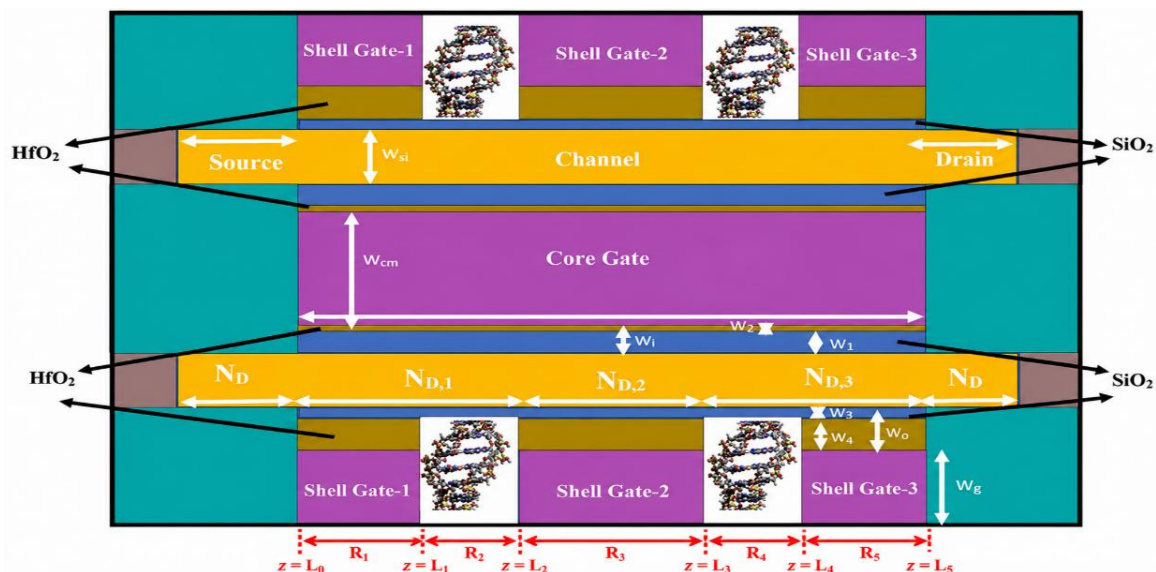


Fig. 2(a). Cut plan view of Bio-NT-JLFET sensor

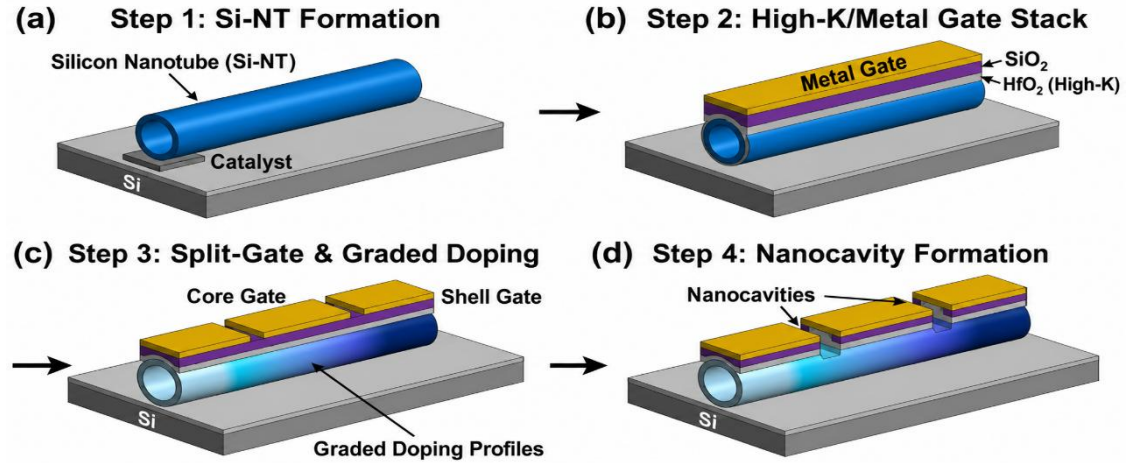


Fig. 2(b). Schematic illustration of the fabrication process flow for the Graded-Channel Split-Gate Bio-NT-JLFET

The fabrication process begins with the formation of a silicon nanotube channel on a prepared substrate, followed by the deposition of high-K dielectrics (HfO_2) and metal patterning to create the split-gate architecture. Subsequently, a graded doping profile is implanted along the channel to suppress leakage, and selective etching is performed to create underlap nanocavities for biomolecule immobilization. It can be seen in Fig. 2 (a) that the channel region bears graded doping concentrations (denoted as N_{D1} , N_{D2} , and N_{D3} , respectively) with N_{D1} being the highest and N_{D3} being the lowest [22]. The numerical values of device parameters like dimensional and doping concentrations of GC-SG-NT-FET-based biosensors, considered during simulation, are listed in Table 1. Note that, the GC-SG-NT-FET structure is compatible with state-of-the-art CMOS fabrication methods. The key process steps, involving Si-NT development, high-k dielectric deposition, metal-gate patterning, ion-implantation for graded channel engineering, and cavity formation through selective etching, are usual part of advanced semiconductor manufacturing. Moreover, the junctionless architecture has the ability to eliminate the need for abrupt source/drain junction formation, simplifying fabrication complexity and reducing process variability. In the light of these characteristics, the device under consideration may be a promising candidate for cost-effective biosensor realization.

The simulation models an ex-situ detection protocol. In this operational mode, the sensor is incubated with the analyte, followed by a washing step with deionized water to remove unbound ions, and subsequently measured in a dry air environment. Consequently, the ionic screening effect (Debye shielding) is minimized, and the background dielectric constant is set to that of air ($K=1$). This approach maximizes the electrostatic coupling between the biomolecule charge and the channel. The environment naturally allows electrical forces to reach much farther, helping the sensor detect the charges of biomolecules more easily and reliably, without any need for extra labels or

indicators. The models executed to perform the simulation of the Bio-NT-JLFET [23] are as follows: Drift-diffusion (DD) charge transport model; High field saturation mobility model (*ccsmob*); the concentration and temperature-dependent Klaassen mobility model for the temperature range from 250 K to 500 K; Lombardi (*cvt*) mobility model to account for the mobility of channel charge owing to carrier to carrier scattering and operating temperature; Shockley-Real-Hall (SRH) with fixed carrier lifetime (1×10^{-7} s) [23].

Table 1. Device dimensions of Bio-NT-JLFET used for simulation

S. No	Parameters	Values
i	Channel length(L)	100 nm
ii	Source/Drain Doping (N D)	10^{19} cm^{-3}
iii	Graded Channel Doping (N_{D1}, N_{D2}, N_{D3})	$5 \times 10^{18} \text{ cm}^{-3}$ $1 \times 10^{18} \text{ cm}^{-3}$ $1 \times 10^{17} \text{ cm}^{-3}$
iv	Inner High K Thickness (w_2)	1 nm
v	Outer High K Thickness (w_4)	3 nm
vi	Inner Oxide thickness (w_1)	2 nm
vii	Outer Oxide thickness (w_3)	1 nm
viii	Thickness of the channel (t_{Si})	5 nm
ix	Core gate radius	5 nm
x	Shell gate-1,2,3 radius	6 nm
xi	Region Length (R_1, R_2, R_3, R_4, R_5)	20 nm, 15 nm, 30 nm, 15 nm, 20 nm

The Bio-NT-JLFET structure was calibrated against an experimentally validated silicon nanotube FET reported by Tekleab et al. [24] as demonstrated in Fig. 3.

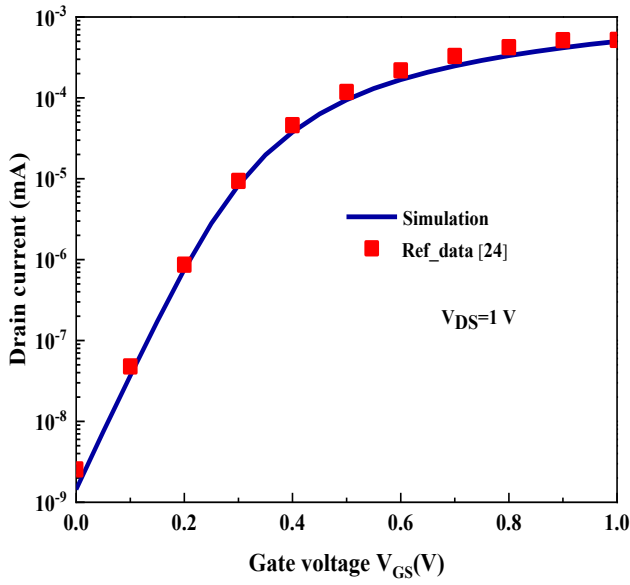


Fig. 3. Calibration graph of drain current I_D vs V_{GS} of NT MOSFET (Ref. [24])

The tuning parameters included effective mobility model coefficients accounting for phonon and surface scattering, SRH carrier lifetimes, and the metal gate work function to match the reference threshold voltage. The extracted I_D – V_{GS} characteristics exhibited close agreement with the published data, with a root-mean-square error (RMSE) $< 5\%$ across the relevant bias range, confirming the accuracy and reliability of the adopted material and transport models. All the temperature-based analyses of the proposed device have been examined with regard to the calibrated results.

3. Results and discussion

This section is dedicated to analyzing and discussing the simulation data extracted from the TCAD device simulator. The electrically-charged biomolecules- single-stranded DNA (*ss*-DNA), double-stranded DNA (*ds*-DNA), and Avian influenza antibody (AI-ab) have been used as a target in the present study. For detection purposes, charge modulation and dielectric principles have been utilized.

The numerical values of dielectric constant (K) and surface charge density (S_q) considered in the subsequent plots have been taken as follows: Air ($K = 1$, $S_q = 0$), AI-ab ($K = 3$, $S_q = -9.6 \times 10^{-8} \frac{C}{cm^2}$), *ss*-DNA ($K = 2$, $S_q = 1.6 \times 10^{-8} \frac{C}{cm^2}$), and *ds*-DNA ($K = 8$, $S_q = 8.0 \times 10^{-8} \frac{C}{cm^2}$) [22].

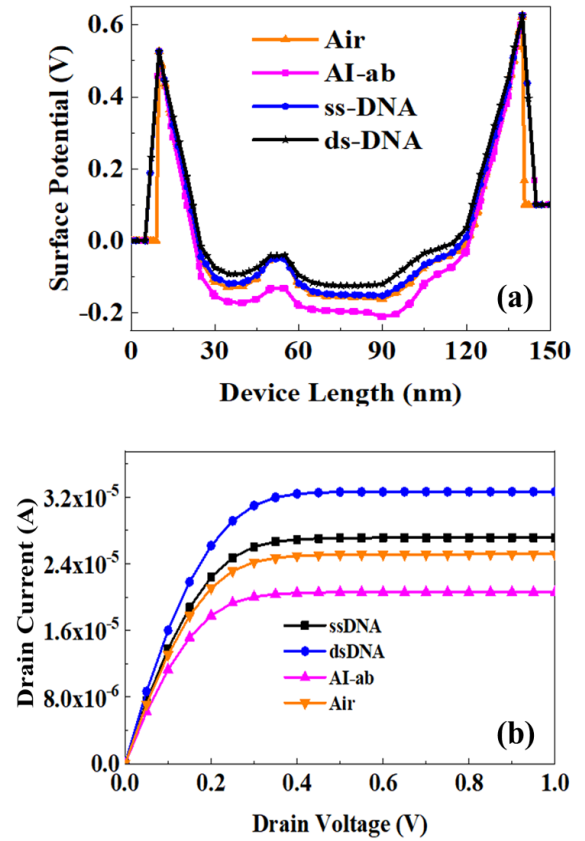


Fig. 4. (a) Surface potential variation along the device length for air and different biomolecules, and (b) output characteristics illustrating the variation of drain current (I_d) with drain–source voltage (V_{ds}) for various biomolecular conditions

The plot of surface potential across the device channel length for the drain-to-source voltage $V_{DS} = 0.1$ V has been given in Fig. 4 (a). Air indicates the absence of biomolecules in the cavities. It can be observed that the surface potential decreases from no biomolecules (Air) to AI-ab biomolecules, where a higher value of dielectric ($K = 8$) provides a maximum uplift in surface potential for *ds* – DNA biomolecules compared to other ones. The introduction of a negative surface charge (S_q) of AI-ab biomolecule in the open cavities results in escalation in flatband voltage, rendering maximum variation in surface potential in the proposed Bio-NT-JLFET device, which culminates in higher depletion in the channel region with elevated gate control.

The effect of *ss*-DNA, *ds*-DNA, and AI-ab biomolecules on drain current (I_d) against drain voltage (V_{DS}) at $V_{GS} = 1$ V is demonstrated in Fig. 4 (b). The higher value of drain current of *ds* – DNA and *ss* – DNA is due to the positive surface charge densities. At $V_{DS} = 0.2$ V, for *ss* – DNA and *ds* – DNA, the drain current value is 6.12% and 24.1% higher as compared to air (no molecule), while for AI-ab, I_d value is 15.7% lower in comparison to air. The downfall of I_d for AI-ab is due to the negative surface charge value. Note that the large variation in I_d at lower values of V_{DS} is good for

distinguishing the different biomolecules. Fig. 5 (a) reflects the plot of the drain current (I_d) against gate voltage (V_{GS}) at $V_{DS} = 1$ V for different categories of biomolecules and air. The left axis indicates the *off*-current (I_{off}) variation, which is the lowest for AI-ab, while the highest for *ds* – DNA owing to different values of K and S_q . The right-hand side axis reflects the same variation of *on*-current (I_{on}) for different biomolecules, but on a linear scale. In comparison with air, for *ds* – DNA and *ss* – DNA, the values of I_{on} are 29.8% and 7.89% higher, respectively, while its value is 18.1% lower for AI-ab as compared to air. The higher variation in I_{on} for *ds* – DNA segregates it from the other biomolecules.

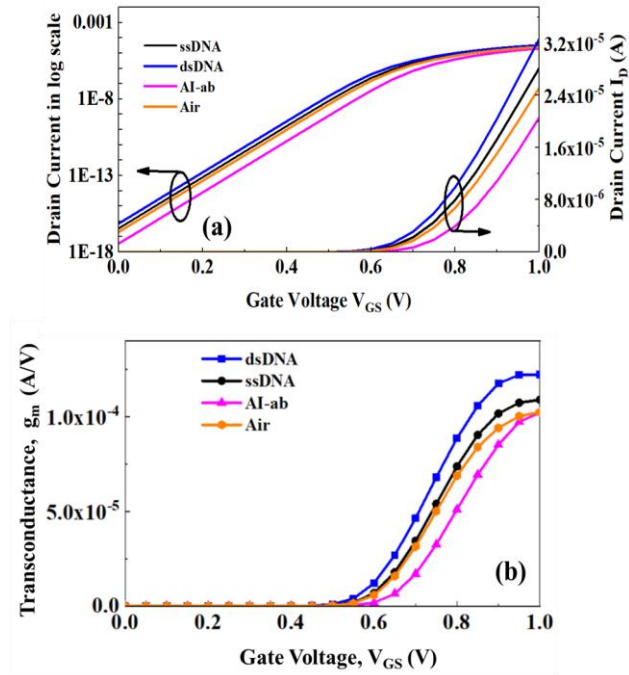


Fig. 5. (a) Transfer characteristics showing the variation of drain current (I_d) with gate–source voltage (V_{GS}) for air and different biomolecules, and (b) corresponding transconductance (g_m) variation as a function of V_{GS} for various biomolecular conditions

Fig. 5 (b) is dedicated to investigating the effect of *ss* – DNA, *ds* – DNA, and AI-ab biomolecules on the transconductance variation (g_m) against V_{GS} at $V_{DS} = 1$ V in the proposed Bio-NT-JLFET sensor. Transconductance (g_m) describes the first-order differential change of I_d against the V_{GS} . Its higher value reflects the good controllability of V_{GS} over the channel. Because of the positive value of S_q , the *ds* – DNA and *ss* – DNA biomolecules show positive drift in transconductance (g_m), while the negative drift in AI-ab biomolecule may be attributed to the negative value of S_q , when inserted in the air-filled cavity of the biosensor. The nature (magnitude and sign of surface charge, S_q) of a biomolecule depends on the type of amino acids it possesses and the pH of its surroundings. In this way, the

behavior of a biological system may be analyzed by variation in threshold voltage, which largely depends on the magnitude and sign of the S_q .

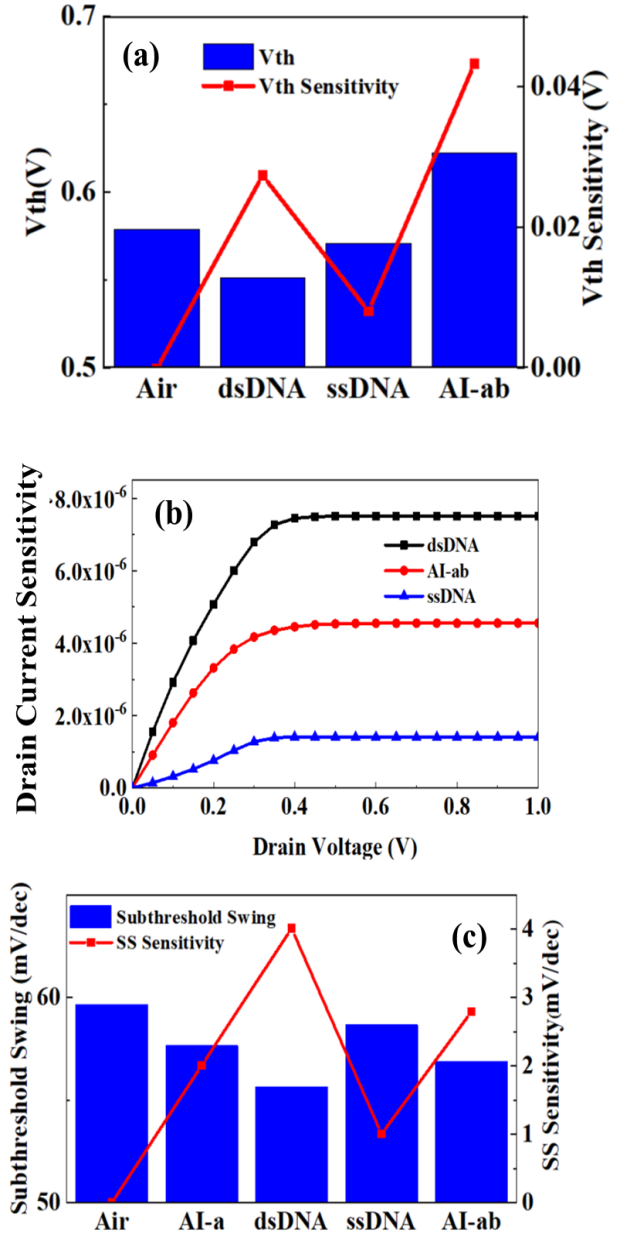


Fig. 6. (a) Variation of threshold voltage (V_{th}) and threshold voltage sensitivity (V_{ths}) for air and different biomolecules, (b) drain current sensitivity (I_{ds}) as a function of drain–source voltage (V_{DS}) for various biomolecules, and (c) variation of subthreshold swing (SS) and its sensitivity for air and different biomolecular conditions

The next figure (Fig. 6 (a)) is attributed to dealing with the variation in threshold voltage (V_{th}) and threshold voltage sensitivity (V_{ths}) of the Bio-NT-JLFET sensor against the presence of three different biomolecules in air-filled cavities. The threshold voltage sensitivity (V_{ths}) is defined as

$$V_{ths} = |V_{thBio} - V_{thAir}| \quad (1)$$

It can be observed that the biosensor carrying *ds* – DNA needs the lowest voltage to be turned on because of the positive and larger value of S_q . Whereas, the negative value of S_q for AI-ab biomolecule renders the highest V_{th} , compared to other sensors. The result also verifies the variation of surface potential observed in Fig. 4 (a). The values of V_{ths} of the Bio-NT-JLFET sensor for *ds* – DNA, *ss* – DNA, and AI-ab are 0.027 V, 0.008 V, and 0.047 V, respectively. Note that the threshold voltage sensitivity has been calculated using Eq. 1, which indicates the minimum sensitivity of the biosensor for *ss* – DNA due to the relative difference between threshold voltages. In Fig. 6 (b), the drain current sensitivity (I_{ds}) of the Bio-NT-JLFET sensor for *ds* – DNA, *ss* – DNA, and AI-ab biomolecules has been graphed. The higher value of the drain current reflects higher sensitivity. The drain current sensitivity is calculated as

$$I_{ds} = |I_{dBio} - I_{dAir}| \quad (2)$$

The biosensor reflects the highest drain current sensitivity for *ds* – DNA while the lowest for *ss* – DNA. The positive value of S_q enhances the charge carrier density in the channel region, giving rise to the drain current in the case of *ds* – DNA filled cavity of the Bio-NT-JLFET sensor. Fig. 10 shows the subthreshold swing and its sensitivity variation for the Bio-NT-JLFET biosensor on various biomolecules when inserted in an air-filled cavity. The subthreshold slope (SS) refers to the variation in the gate voltage required to alter the drain current by one decade (a factor of ten) when the device is operating in the subthreshold region. The SS with a sharp slope indicates the fast ON/OFF transition performance of the device [24]. The Bio-NT-JLFET biosensor shows an increasing order of SS value for *dsDNA*, AI-ab, AI-a, and *ssDNA* biomolecule when inserted in an air-filled cavity. The subthreshold swing sensitivity of the Bio-NT-JLFET biosensor is defined as

$$SS_s = |SS_{Boi} - SS_{Air}| \quad (3)$$

Based on the relative difference between the SS value of biomolecules and air, the Bio-NT-JLFET biosensor demonstrated the minimum sensitivity (SS_s) value for *ss* – DNA biomolecules when inserted in an air-filled cavity at 1.1 mV/dec. Fig. 7 (a-c) depict the sensitivity variation against VDS across a broad temperature range (250 K to 500 K). It is noted that while biological analytes typically denature above 320–340 K, the analysis extends to 500 K to evaluate the semiconductor device's thermal stability.

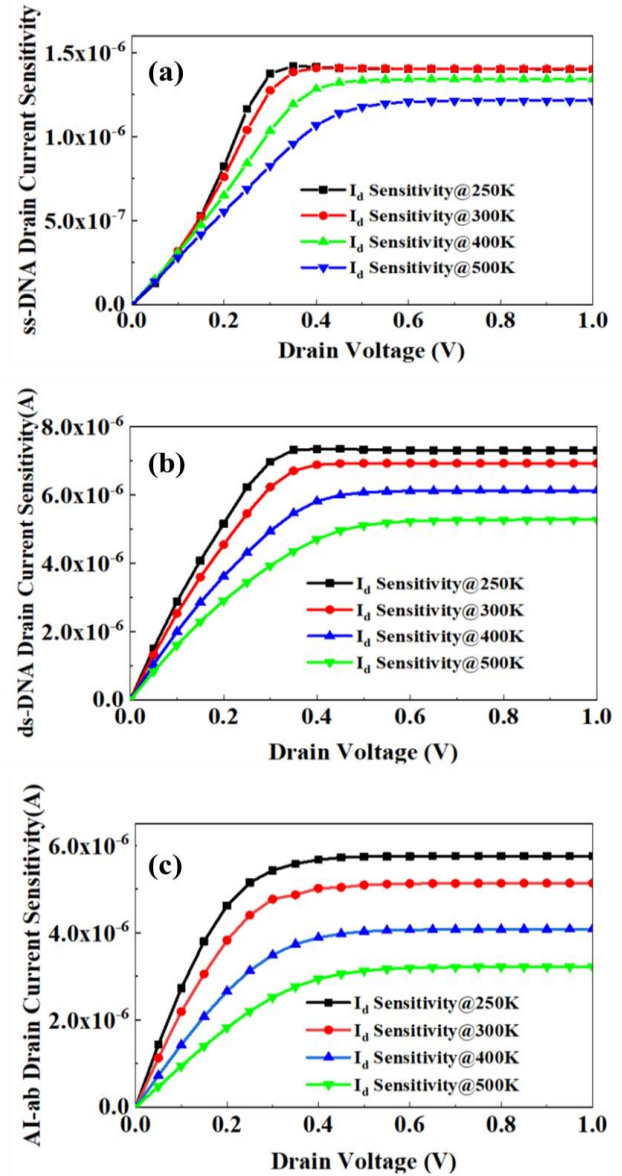


Fig. 7. Plot of temperature-dependent drain current sensitivity characteristics of the proposed Bio-NT-JLFET biosensor: (a) variation of drain current sensitivity for *ss*-DNA, (b) variation of drain current sensitivity for *ds*-DNA, and (c) variation of drain current sensitivity for AI-ab biomolecules as a function of drain-to-source voltage (V_{DS}) over the temperature range of 250–500 K

The drain current sensitivity to temperature variations is a complex interplay of threshold voltage shift, carrier mobility variation, carrier density change, device geometry, and material properties. Fig. 7 (a) covers the variation of the drain current sensitivity of the Bio-NT-JLFET sensor against V_{DS} at various temperatures when the *ss* – DNA biomolecule is inserted in an air-filled cavity. On increasing the temperature, the drain current sensitivity (I_{DS}) goes down due to the degradation of carrier mobility caused by increased phonon scattering. Fig. 7 (c) and (d) demonstrate the Bio-NT-JLFET sensor drain current sensitivity variation against V_{DS} at various temperatures when the *ds* – DNA and AI-ab biomolecules

are inserted in an air-filled cavity. Variation in sensitivity at different temperatures follows the same pattern as in Fig. 7 (a).

It has been observed from Fig. 7 (a-c) that increasing the temperature from 300 K to 500 K at $V_{DS} = 1$ V, the fall in drain current sensitivity (I_{DS}) for $ds - DNA$ is 23.8%, $ss - DNA$ is 13.5%, and $AI-ab$ biomolecule is 37.3%. The next plot (Fig. 8 (a)) discusses the variation of the I_{on} sensitivity of the Bio-NT-JLFET sensor for the temperature range from 250 K to 500 K for $ss - DNA$, $ds - DNA$, and $AI-ab$ biomolecules. The sensitivity has been found to be decreased with increasing temperature owing to the complex effect of device geometry, channel doping, and mobility variation of the device with temperature.

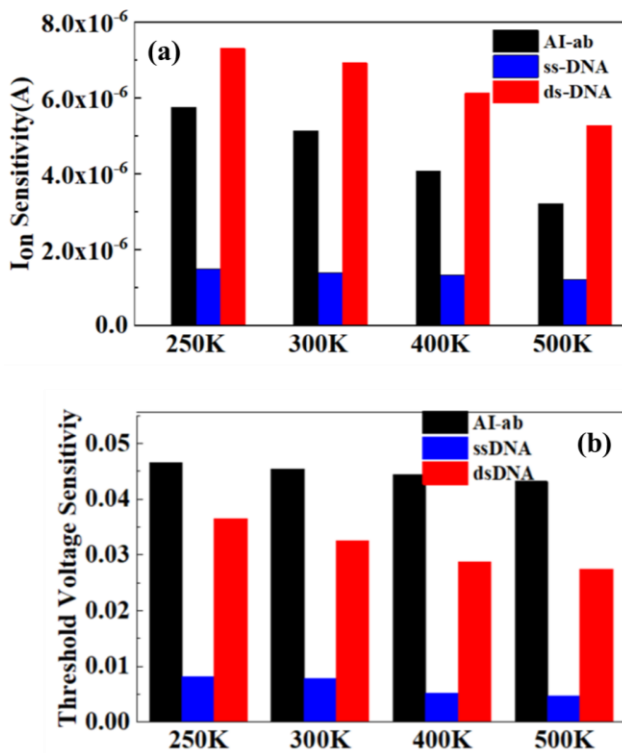


Fig. 8. Plot of temperature dependence of key sensing metrics of the Bio-NT-JLFET biosensor: (a) variation of on-current sensitivity and (b) threshold-voltage sensitivity for different biomolecules (AI-ab, ss-DNA, and ds-DNA) across the temperature range of 250–500 K

Fig. 8 (a & b) depicts the plot of on-current sensitivity and threshold voltage sensitivity (V_{th}) of the Bio-NT-JLFET against temperature variation ranging from 250 K to 500 K. The threshold voltage sensitivity has been found to be decreased with increasing temperature. With increasing temperature, V_{th} shifts downward ($|\Delta V_{th}|$ often reduces at a fixed bias), while mobility typically degrades due to phonon scattering. Although promising for dry and low-ionic sensing applications, the present study does not include electrolyte screening and pH-dependent variations in surface charge. In physiological buffers with short

Debye lengths, the sensor response may be reduced owing to ionic shielding. Incorporating ionic-strength-dependent electrostatics and surface chemistry-based charge modulation will be considered in future extensions to ensure practical implementation in biofluid conditions. Compared with Tunnel FET and 2D-material FET-based biosensors [25, 26], which offer much larger current sensitivities, often rely on different transduction metrics, liquid-gated operation with screening mitigations, and require complex fabrication, the Junctionless nanotube FET provides a simpler, CMOS-compatible structure with reliable and stable charge sensing. This makes the Bio-NT-JLFET a practical and scalable choice for label-free biomolecule detection.

4. Conclusion

In this paper, various sensitivity analyses of the Bio-NT-JLFET biosensor have been performed for detecting the *AI-ab* and *DNA* biomolecules. The Bio-NT-JLFET employs a silicon nanotube FET with graded channel doping for operation in Junctionless accumulation mode (JAM). The Bio-NT-JLFET biosensors are designed for the effective detection of *DNA* and the avian influenza (*AI - ab*) virus. It has been found that the graded Junctionless NT FET structure, along with two open cavities, efficiently improves the sensor's performance, particularly in terms of threshold voltage sensitivity (V_{th}) and drain current sensitivity (I_{ds}). The performance of the device got ameliorated with the combination of graded channel, core-shell gate, and shell gate, and triple-metal work-function engineering with enhanced control of gate terminal over the charge carriers in the channel region. This study shows that the graded-channel split-gate Bio-NT-JLFET design can clearly detect different biomolecules using both current and voltage signals. Among them, the *AI-ab* biomolecule causes the largest change in threshold voltage, about 47 mV, while the *ds-DNA* molecule gives the biggest increase in current, around 29.8% when the voltage $V_{DS} = 1$ V and $T = 300$ K. The device exhibits robust electrical switching up to 500 K, indicating high thermal stability for the sensor hardware. However, peak sensitivity is achieved at lower temperatures (300 K), with a sensitivity degradation of 23.8% for *ds-DNA* as temperature increases to 500 K. This confirms the device's suitability for room-temperature ex-situ applications with a high tolerance for thermal stress. The calibrated TCAD simulations show that the device simulation works correctly in physical terms and is well-suited for biosensing without labels, in low-ionic conditions, and with temperature consideration. It is to be noted that the detection of *AI-ab* is of paramount importance in early veterinary related diagnosis. Rapid and label-free identification of *AI-ab* biomarkers may help primary disease screening, thereby reducing the risk of viral transmission, with timely implementation of preventive measures. In this way, the proposed Bio-NT-JLFET biosensor may prove to be of substantial potential

for future point-of-care and widespread diagnostic applications.

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