Investigation into MoTe$_2$ Based Dielectric Modulated AMFET Biosensor for Label-Free Detection of DNA Including Electric Variational Effects

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Abstract
Due to the limitations of Silicon, Transition metal dichalcogenides (TMD) based biosensors are popular in recent times. In the TMD family, Molybdenum telluride (MoTe$_2$) is being studied a lot for different biosensing applications. However, for DNA detection using TMD-based DMFET, the effect of the electrical variations in DNA has not been studied before. Also, the impact of DNA-Electrode interaction on the transducer level of DMFET is yet to be studied. In this article, we have proposed a Molybdenum telluride (MoTe$_2$) based Accumulation Mode Field Effect Transistor (AMFET) for possible dielectric modulated biosensing application. The study is focused on DNA detection including the electric variations of DNA due to surface interaction. We have done a circuit-level analysis of the proposed structure for having deeper insights into its performance under various DNA orientations in the nanogap. We have also presented a benchmarking to highlight the superior sensitivity of the proposed structure ($\Delta V_{th} = 700$ mV at $K = 8$). The impact of back-gate bias is also included. We have obtained significant variation of threshold voltage shift for different orientations in the proposed structure suggesting a strong impact of electrical variations in DNA in the biosensing performance of MoTe$_2$ AMFET.

Keywords
Deoxyribose nucleic acid (DNA) · Biosensor · DMFET · Transition metal dichalcogenides (TMD) · Circuit analysis

1 Introduction
To serve a larger spectrum of humanity, biosensors have been evolved and enabled for label-free detection in the arena of agriculture, medicine, ecological surveys, food industry, etc. [1, 2]. The reason behind the popularity of label-free biosensors is its immediate response capability for bio-analyte identification without complex probe arrangements. A large amount of research has been done on biosensors for the detection of proteins, viruses, and DNA [3–9].

Ion-sensitive Field Effect Transistor (ISFET) was the most trivial biosensor which was proposed a long way back. Its performance is exceedingly well in the case of charged molecule detection. However, the neutral molecule detection is not so overwhelming in the case of ISFET [10]. In comparison to (ISFET), which on the charge interaction effect, Dielectric Modulated Field-Effect Transistors (DM-FET) are capable of detection of non-charged biomolecules & thus have a wider perspective. Widescale research has been carried on DMFET biosensors for years [3–9]. While some studies focus on increasing the sensitivity of DMFET biosensors, others emphasized unveiling the underlying physics of such label-free biosensors. FET biosensors work on two principles i.e., (i) charge interaction and (ii) dielectric constant modulation effects. The biomolecules immobilize inside the craved cavity in DMFET, modulates the effective oxide capacitance, and consequently, effective dielectric coupling between gate & channel varies. Changes in electrical properties quantify the sensitivity for label-free detection for both neutral & charged molecules viz. biotin-streptavidin & DNA [3–9]. The approach played a vital role in making DMFET a widely explored structure for label-free biosensors with inherent advantages like higher sensitivity, power consumption, higher scalability & fabrication simplicity.

In keeping with the advancement in semiconductor technology, the research aims for smaller FET Biosensors with a higher density as well as higher sensitivity [11].
cavity thickness touching sub-10 nm regime, several research works are focusing on FinFET & nanowire FET for detection of DNA, Proteins, viruses, etc. [12]. At scaled dimensions, Silicon-based FET biosensors are found to be inadequate to provide crucial sensing performance metrics. After the isolation of graphene in 2004, research works in ultra-scaled DMFET biosensors have been flourishing based on TMD (MoS2, MoSe2, WS2, WSe2, etc.) material. The atomic thinness along with a higher surface/volume ratio result in a stronger response to surface adsorption phenomena & makes TMDs the best-suited material for sensing applications [12]. In addition, denser active surface sites, broader & tenable electronic properties (due to layer-dependent band structure), enhanced selectivity to specific analytes & extremely high sensitivity opens up the endless opportunity for TMD-DMFETs to be next-generation scaled biosensors for particularly for healthcare applications [13]. For detection of DNA (exceedingly important for medical research, cancer diagnostics, forensics, etc.), the traditional process of preamplification & optical detection (i.e., measurement of fluorescence intensity of labeled strands, etc.) demands specialized infrastructure & human resources [13]. Here, TMD-based DMFETs for DNA electrochemical detection emerged as a cost-effective, speedy detection method that too with a much lower detection limit in the range of Femto- or attomolar. Detection of different DNA strands by 2D-DMDFETs has been explored by researchers viz. Jin et al. on Dengue DNA using graphene oxide and wrapped SiO2 particles. Zhang et al. for tumor DNA. using exfoliated MoS2, Checkin et al. for HPV DNA, using graphene oxide (prGO)/MoS2 composite [12]. The linear range & detection limit of the last cited research has been reported 3.5–35.3 pM, and 1.75 pM respectively [14]. Very recently, Hwang et al. has reported crumpled graphene 2D material-based FET biosensor with ultrasensitive detection of DNA and RNA molecules with a significantly low limit of detection (LOD) [15].

For more advanced applications, MoTe2 is being considered suitable material for biosensor 2D material-based DMFETs due to a smaller bandgap, lower thermal conductivity with a higher Seebeck coefficient when compared with other 2D materials [16]. In addition, MoTe2 showed unique & viable properties in the TMD family in terms of growth, bandgap engineering, carrier injection, etc. [16]. A high ratio of on/off current with a low subthreshold swing made the MoTe2 one of the most suitable candidates for a device, logic circuits, optoelectronics as well as sensing application. Recently, Feng et al. reported a high sensitivity of the molybdenum ditelluride (MoTe2) sensor with a significantly improved recovery rate [17]. In literature, it has been widely reported that experimental results underperform theoretical prediction due to irregular arrangement of biomolecules in the cavity, steric hindrance effect, partial hybridization & weak binding possibility, probe-arrangement variability, etc. [18]. As these lead to significant performance degradation, the minimum detection limit becomes a vital performance parameter in addition to sensitivity in the scaled regime.

In this manuscript, we have proposed a MoTe2 based Accumulation Mode Field Effect Transistor (AMFET) for biosensing application. In the following sections, we have mainly focused on understanding the MoTe2 based biosensor performance for DNA detection. A general sensitivity analysis of the proposed structure has been presented at the beginning and the device performance has been benchmarked with previously reported DM-MOSFET studies. The impact of the interaction between the DNA strands and gate electrodes is one of the experimentally validated sources of variability. The effect of electrical variations in DNA due to interaction with an electrode on the performance of biosensors is studied in detail. We have also considered the effect of irregular orientations of DNA on biosensor performance in form of case studies. The study has been conducted on both device and circuit levels.

### 2 Device Structure and Simulation Methodology

MoTe2 Accumulation Mode Field Effect Transistor is proposed as the device under analysis. The channel height is taken to be 10 nm. The channel length is 35 nm. Source and Drain extensions are 20 nm each. A SiO2 box is considered of height 10 nm as well. Cavity height is kept constant at 10 nm. A small portion of SiO2 is considered for the immobilization of biomolecules. The top and bottom gates are considered to be Gold(Au) with work-function 5.1 eV. Source and Drain regions are doped heavily with a concentration of $10^{20}$/cc (n-type), while the channel region is doped very lightly with a concentration of $10^{16}$/cc (n-type). The detailed schematic of the device structure can be found in Fig. 1.

The proposed AMFET has been simulated using SILVACO TCAD [19]. The study has been done on two levels. The first is the device level analysis and the second one is the circuit level assessment. For the first set of studies, we have adopted Fermi-Dirac statistics as the main carrier statistics. When the device is in thermal equilibrium, carriers seem to obey Fermi-Dirac statistics with the semiconductor lattice [19]. For mobility, we have adopted the Concentration-dependent mobility model and Field Dependent mobility models for capturing mobility variation under high electric fields. Lombardi mobility models have been taken into consideration to account mobility degradation effect at the inversion layer due to the high surface scattering phenomenon. Shockley-Read-Hall recombination model is included. Since the source and drain are highly doped regions, there will be significant band bending in these regions due to high doping. To include this in calculations, Bandgap narrowing effects
have also been taken into account. Quantum effects have not been included as quantum confinement plays a major role in the sub-10 nm domain and this work has been done with a channel thickness of 10 nm. All the models mentioned above along with a drain voltage of 0.5 V (for low energy operation) are deployed to obtain the DCIV characteristics of the proposed architecture and other device-level parameters like threshold voltage etc. Our simulation setup is well calibrated and shown in the Supplementary File. For the second set of studies i.e., circuit-level assessment the proposed structure along with the model mentioned above are fused into the MIXED MODE package of SILVACO ATLAS [19]. Both DC and Transient simulations using the above structure are done for different levels of analysis to explain the device behavior in detail.

The proposed structure is intended to perform as Dielectric Modulated Biosensor. For simulating different biomolecules, the electrical properties of the nano-particles are taken into considerations. The cavity dielectric constant is varied regarding different biomolecules (neutral). For charged biomolecules, the effective charge effect is being considered in form of the interface traps at the cavity and channel interface. For circuit simulations as well, this methodology has been followed.

3 Benchmarking

In this manuscript, we have mainly focused on understanding the biosensor performance for detecting the DNA biomolecules, which are charged in nature. However, to establish that this proposed biosensor offers superior sensitivity for neutral biomolecules as well, we have presented a benchmarking with the already reported Silicon-based biosensor studies in Fig. 2. The data for low dielectric constant ($K = 2$) and high dielectric constant ($K = 8$) based biomolecules are presented in the above figure. We can witness that the proposed MoTe2 based AMFET outperforms other Silicon-based biosensors for both high and low dielectric constant-based biosensors. It is to be noted that for benchmarking purposes, we have not considered the DNA variability as the previous studies do not include this factor.

4 Results and Discussion

In the following sections, we have presented various analyses both on the device level and circuit level for understanding the DNA detection capability of the proposed structure and also the way, performance is hampered by variability sources. The sensitivity of a parameter $Q_1$ is calculated following the equation:

$$S_{\text{sen}}(Q_1) = \left| \frac{Q_1(\rho = c) - Q_1(K = 1)}{Q_1(K = 1)} \right|$$

The term $K = 1$ stands for the bare device condition i.e., the cavity is not filled with any biomolecules or linkers. $\rho$ stands for the charge density of the biomolecules. $\rho = c$ means a charge of a particular molecule is $c$. Another important study is selectivity. The selectivity of a parameter $Q_1$ for one condition $u_1$ with respect to other $u_2$ is calculated by the following equation:

$$S_{\text{sel}}(Q_1) = \left| \frac{Q_1(u_1) - Q_1(u_2)}{Q_1(u_2)} \right|$$
These equations are used for all the sensitivity and selectivity study done in the upcoming sections. For the study without DNA-Electrode Interaction Effect, we have considered conventional approach while for that including DNA-Electrode interactions we have relied on experimental setup as mentioned in [20–22].

4.1 DNA Detection without Considering DNA-Electrode Interaction

As mentioned in [23], a monolayer of ssDNA (single-stranded DNA) is genetically coded into the cavity which serves as the base for DNA hybridization. This base layer is particularly efficient in binding with specific target DNA strands which results in high selectivity in target detection. This step is shown in Fig. 3b. The DNA next hybridizes with this base layer after immobilization to form dsDNA. After some time, the number of full hybridized dsDNA increases, and we get the cavity filled with DNA biomolecules. To have a deeper insight into the transition from partial to complete hybridization of DNA, we have analyzed one of the intermediate stages of partial hybridization as shown in Fig. 3c. Figure 3d shows the completely hybridized cavity. From a simulation point of view, each of the stages as shown in Fig. 3 and discussed above is associated with definite dielectric properties. The details of the dielectric constant and charge of each layer are provided in the Supplementary File. With drain voltage ($V_{ds}$) of 0.5 V and back-gate at zero bias condition, the top-gate bias ($V_{TG}$) is varied up to 4 V to get Current-Voltage characteristics of the proposed biosensor during the four stages of DNA detection. For bare device, i.e., when a cavity is filled with air, the gate capacitance is the least. Consequently, electric coupling between the gate electrode and semiconductor lattice is worst. A smaller number of electrons are pulled towards to surface to form the conduction channel which results in a very low current. Also, less gate controllability over the channel causes comparatively higher OFF current as can be seen in Fig. 4a (Dark Blue). Now as the ssDNA base layer is formed in the cavity, the channel conduction increases due to improvement in gate-channel coupling which presents a better current profile. As we transit from partial to complete hybridization, the effective gate capacitance increases which result in higher $I_{ON}$ and lower $I_{OFF}$. All features can be seen in Fig. 4a. From a biosensor perspective, the sensitivity and selectivity of different stages are important. As can be observed in Fig. 4b, the current sensitivity is very high for the Completely hybridized stage in comparison to the bare device. The max Current sensitivity is about $10^5$ for the Completely hybridized stage, which is $10^4$ and $10^2$ for partially hybridized and ssDNA stages respectively. To differentiate between the different stages, we have analyzed the selectivity as well. From Fig. 4c, we find the selectivity of the fully hybridized stage is approximately $7 \times 10^4$ and 100 with respect to the ssDNA stage and partially hybridized stage and a reduction of almost 61% can be seen in between these two cases. Thus, we may conclude that MoTe$_2$ based AMFET can differentiate between levels of DNA detection with a good level of precision. The Threshold sensitivity is also much higher for the completely hybridized DNA stage (approx. 0.29) in comparison to 0.14 and 0.21 in the case of ssDNA and partially hybridized dsDNA respectively.

4.2 DNA Orientation Effect and Implementation Methodology

The immobilization of DNA plays an important role in biosensing performance. Specifically for DNA biosensors, the interaction of DNA with the surface is a vital factor for DNA genome sequencing or biosensing. Several kinds of research have been carried out over the years to understand the characterization of DNA when it comes in contact with the surface [20–22]. Among others, some noteworthy problems associated with single-molecule systems like DNA are their orientation, biological affinity, electrode surface status under microenvironment, and so on. In addition to these, previous studies [21, 22] have reported alteration in DNA electrostatic characterization also controls the nature of interaction with the surface. As mentioned in [21, 22], the way DNA interacts with gold surfaces is greatly affected by the electrostatic variations, precisely the orientation of DNA present on them. If the applied bias exceeds the potential of zero charges (PZC) for any electrode, then we will witness such variability in DNA orientation [20, 22]. The potential of zero charges is defined as the potential value at which net charge density comes to zero on the electrode surface. This PZC depends on several factors like the concentration of the solution, the microenvironment of the electrode, and so on. Hence, the concept can be extended to any electrodes, provided its surface interaction with
DNA is well verified. Since we have experimental evidence for the gold electrode (PZC = 0.26 V), we have considered gold to be a gate electrode in our simulations as well. In Fig. 5, we have presented a schematic of different orientations a DNA can have in course of its interaction with the gold surface. The brown arrows show the electric force of attraction of the gate electrode. Other forces working on the single-molecule system, DNA are the mutual repulsive force (as DNA is negatively charged in nature) and the repulsion from the majority carriers in the semiconductor lattice (as it is n-type; for p-type there would have been attraction). For our simulation purpose, we have considered a self-assembled monolayer mainly consisting of thiol linkers. The value of the relative dielectric constant of the layer is considered '$2$' [20]. The length of DNA strands is considered to be 7 nm and has a relative dielectric constant of 8. The inter-strands distance is 2 nm. It has been reported in [20, 21], that double-stranded fully hybridized DNA strands can be considered to...
be rigid rods with homogenous charge distribution over them. Hence, we have considered, if the COM of the DNA leans by an angle of concerning the SAM layer, both the strands of DNA will incline at the same angle. This approximation is a valid one as reported in [20]. In the following sections, we shall see the effect of this variability in detail from device and circuit perspectives.

4.3 DNA Orientation Effect – Device Level Perspective

A large variation in the potential profile and electron distribution can be seen for DNA oriented vertically and horizontally in Fig. 6. This level of variation has significant effects on the device performance as can be seen in Fig. 7. In Fig. 7, we have presented the ‘Current Sensitivity’ and ‘Threshold Sensitivity’ of the proposed MoTe₂ based architecture which are important from biosensor performance. The Id-Vgs profiles for each of these different orientations are provided in Supplementary Material. In Fig. 7a, the sensitivity of the vertical position is about $6 \times 10^3$ while that for horizontal is almost 1. Sensitivity above 1 is considered appreciable. The reason for such a low sensitivity of horizontal position is that the amount of gate coupling increased due to the 2 nm addition of DNA strands is not sufficient to cause the appreciable amount of current conduction. Sensitivity for 45° inclination of DNA strands is $1 \times 10^3$ which is 6 times less than that for vertical position. From Fig. 7b the threshold sensitivity is very high for vertical or near-vertical positions but they decrease as DNA inclines more and more towards the SAM layer. Threshold sensitivity for a vertical position is about 4 times more than that of a horizontal one. Hence, we can safely say the orientation of DNA has a significant impact on biosensing performance.

4.4 DNA Orientation Effect – Circuit Level Perspective

4.4.1 DC Simulations

For the circuit level assessment, we have considered a resistive-load-based NMOS inverter as shown in Fig. 8. The source end of the biosensor is always kept grounded. The drain is connected with the resistive load of 100KΩ which in turn is connected to a DC voltage source of value 2.5 V. In the biosensor architecture, there are two nodes at which inputs are given. The top gate is the main node here. V_{TG} is considered to be a constant value for DC simulations and it is in form of a pulse for Transient simulations. The back gate voltage is constant at a definite value for both DC and Transient simulations. The output is taken out from the drain of the biosensor. However, the back-gate bias value has been changed and its effect on biosensing performance is discussed in the next sections. For DC analysis, we have presented the biosensing performance in terms of ‘Voltage Transfer Characteristics’, ‘Output Voltage Selectivity’, ‘Input-High (V_{IH})’, and ‘Input-Low (V_{IL})’ voltage selectivity.

The voltage transfer characteristics in Fig. 9a show the inverter performance with vertical orientation is better than all other orientations. The reason is the better opportunity of gate coupling in case of vertical orientation as can be seen from the above device level studies. To have insights on the comparative performance of different orientations, we have
plotted the output voltage selectivity curve with respect to the vertical orientation in Fig. 9b. We can witness a high selectivity for the horizontal orientation. A peak selectivity of 50 for horizontal position and 32, 20 and 5 for 15°, 30° and 45° respectively. In Fig. 9c, in terms of selectivity of Input-High (V_{IH}) and Input-Low (V_{IL}) voltages, we can see 0.5 (average) selectivity for the horizontal position with respect to the vertical position. Thus, we can conclude that the MoTe₂ based biosensor can successfully differentiate between the horizontal and vertical orientation of the DNA in the nanocavity. Also, it is important to notice that the DNA interaction plays a very important role in biosensing performance which is presented further in the next sections.

4.4.2 Transient Simulations

In the previous paragraph, we have studied the circuit performance in case DC simulation. In this paragraph, we shall be discussing the transient simulation results. In Fig. 10a, the Input-Output voltage curve is plotted. It can be seen that there is a distinctive difference in the characteristics for horizontal and vertical positions. The difference is so significant that it can easily be measured with any voltmeter. The same can be said for the Current-Voltage profile as well in Fig. 10b. The vertical position shows a very steep switching whereas the horizontal position due to lack of enough channel controllability of the gate has very slow switching. Any intermediate orientation angle is to have characteristics in between the red and blue curves shown in Fig. 10a and b. Thus, from transient simulations as well one can easily decipher the orientation of the DNA in nanocavity. The propagation delay has been calculated by considering the average of rise-time delay and fall-time delay.

For propagation delay selectivity, the delay in the case of vertical orientation is compared with that of other orientations. The propagation delay of the inverter for all orientation angles are is shown in Fig. 10c and the selectivity of them with respect to vertical orientation is shown in Fig. 10d. The delay
decreases from horizontal to vertical orientation suggesting faster switching in case of higher orientation angles. For horizontal orientation, the cavity is mostly covered with air. As a result, the gate coupling could not form the inversion channel for easy conduction of current in the AMFET. As a result, significant delay can be noticed in this case. However, in the case of Vertical Orientation, the delay decreases by almost 1.3 ns due to improved channel formation. Due to this high variation of gate coupling effect in both these cases, we see a selectivity of about 20 for the horizontal case with respect to the vertical one. The complete transient profile for Output Voltage and Output Current can be seen in Fig. 10e and f respectively, where both switching and delays can be seen.

4.5 Impact of Back-Gate Bias

For all the analysis, until this point, the back-gate bias has been considered to be zero. However, for a holistic study of DNA detection with MoTe₂ AMFET, the study for the effect of back gate bias is really important. In Fig. 11a it can be seen that with $V_{BG} = 1$ V, the electron concentration in the channel increases and becomes more uniform when compared to Fig. 6c and d. The reason is that higher back-gate bias causes more majority carrier influx into the channel which results in higher electron concentration near the surface. Also, it reduces the impact of top-gate over the channel as for both horizontal and vertical orientation, electron distribution is more or less the same. Thus, qualitatively, it reduces the sensitivity of the proposed biosensor. For quantitative explanation, we can see Fig. 11e and f. In the previous section, we discussed the significant difference for Output Voltage and Output Current between horizontal and vertical cases that can help us to determine the nature of DNA interaction with the surface. But with increased back-gate bias, both switching and delay of the adopted NMOS architecture are nearly equal due to the availability of more electrons at the surface and reduced impact of top-gate. As a result, we cannot understand the DNA interaction nature for increased back-gate bias. Transient Current and Voltage Outputs are provided in Fig. 11c and d. The impact of back-gate bias on the biosensor sensitivity from a device level is also studied and provided in detail in the Supplementary File (Fig. S3,S4,S5). A degraded sensitivity is observed from device-level perspective as well.
4.6 Variation of Biosensor Performance for Irregular Orientation of DNA

In the previous section, we have analyzed the effect of DNA orientation based on an implicit assumption that all the DNA strands have the same level of orientation. Because of this reason, we have considered the DNA layer-end to be a flat line. In a real-time experiment, many factors cause hindrance in achieving this flat line of DNA layer. The steric hindrance causes the DNA to be non-uniformly placed which will prevent all the DNAs to be under the same influence of gate voltage. Again, the initial layer of DNAs near the electrode shields the gate voltage for other layers. These factors together cause a non-uniform degree of rotation for all DNAs. Thus, for having deeper insights into these non-uniform orientations, we have done some case studies. In Case A, Case B, and Case C, the maximum height of the DNA layer is 7 nm while for Case D, Case E and Case F is 3 nm. For Case A and D, the peak DNA layer is near the source while for Case B and E, the peak is near to the middle of the channel and for the last set of
cases, the peak is near to the drain effect. In Fig. 12(TOP) the potential profile of the channel for each of the cases is shown. It can be seen that the potential profile is highly varying in the case of a larger peak height of the DNA layer in comparison to that with a smaller peak height. It can also be witnessed from the above plots that as the peak position changes from the source end to the middle to drain end, the potential profile changes considerably, especially for larger peak height. Out of these three positions, when the peak is in the middle, the potential distribution over the channel is more uniform than the others. Thus, we expect the device to have better performance in this case in comparison to the other ones. Again, when the peak position is near the drain, the surface potential is higher than the rest of the cases. Higher surface potential suggests higher electron concentration near the surface which refers to higher current conduction. It can also be noted that for lower peak height, the peak position has very less effect on the potential profile. The reason can be attributed to the less channel controllability of the gate.

In Fig. 12(BOTTOM), the threshold voltage selectivity has been plotted with different peak positions for different peak heights. As discussed above, for low peak height i.e., the cases
with $t_{\text{peak}}$ equal to 3 nm, the threshold voltage selectivity change for different peak positions is very nominal as the value centers around 0.1. The selectivity variation increases as we move towards higher peak heights. For $t_{\text{peak}}$ equal to 5 nm, the selectivity is 0.13, 0.11, and 0.198 for $L_{\text{peak}}$ equal to 5 nm, 15 nm, and 30 nm. This variation increases much higher in case $t_{\text{peak}}$ equal to 7 nm. The reason is for higher peak height, a slight change in DNA position causes a large amount of gate capacitance changes which is lower in the case of low peak height. From Fig. 12(BOTTOM), it should also be noted that the selectivity is higher for drain-side peak position as the higher electron concentration in the surface in this case results in higher current. This has been supported by the potential distribution in Fig. 12(TOP) (c).

5 Conclusion

In this manuscript, we have proposed a MoTe$_2$ based AMFET for dielectric modulated biosensing performance. We have
specifically studied DNA detection including the DNA-
Electrode interaction effects. For our study, the DNA interac-
tion with gold surface has been taken into consideration which
is experimentally proven. We have noticed a significant
change in the biosensing parameters due to DNA orientation
within the cavity. For a highly sensitive DNA biosensor like
MoTe$_2$ based, it is crucial to consider the DNA interaction
with the electrodes in a microelectronic environment. We
have mainly focused on the impact of such variability from
a circuit-level perspective. The impact of back-gate bias on the
device performance has also been analyzed in detail. Lastly,
we have considered a much orientational structure that has a
high probability of occurring in real-time experiments. For
these cases, the performance of the proposed structure has
been analyzed. Hence, this study can be considered as impor-
tant in the view that it presents the detailed effect of DNA
surface interactions on biosensing performance of highly sen-
sitive MoTe$_2$ based biosensors, which are considered to be
next-generation biosensors.

6 Future Scope

The main focus of the work is to introduce a two-dimensional
material (MoTe$_2$) based AMFET Biosensor for label-free
DNA detection including the non-ideal effect of DNA orienta-
tion within the cavity. As a future scope of the work, we
propose the following:

- One of the important things for a biosensor is Response
time. Thus, the work can be extended to analyze the var-
ation of Response time during DNA detection.
- Another important future work can be preparing an ana-
lytical model of the proposed biosensor system and trans-
fer it to Verilog-A. This will help us to perform a read-out

circuit-based analysis.
- Due to the aggressive scaling of semiconductor devices,
biosensors are also needed to be scaled to integrate a large
number of them in a small area but at the same time,
boosting the sensing capabilities. This motivates us to
pursue the work with different layered MoTe2. The cur-
cent work is based on bulk-MoTe2, it can be extended to
mono and bi-layered MoTe$_2$.
- As mentioned in the previous point, one of the better ar-
rangements of biosensors can be array configuration. Thus,
it will be interesting to explore array level perfor-
ance of the proposed biosensor.

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Declaration

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