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To cite this article before publication: Chawina De-Eknamkul *et al* 2019 *2D Mater.* in press <https://doi.org/10.1088/2053-1583/ab5ae2>

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MoS₂-Enabled Dual-Mode Optoelectronic Biosensor Using a Water Soluble Variant of Mu-Opioid Receptor for Opioid Peptide Detection

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Received xxxxxx
Accepted for publication xxxxxx
Published xxxxxx

Abstract

Owing to their unique electrical and optical properties, two-dimensional transition metal dichalcogenides have been extensively studied for their potential applications in biosensing. However, simultaneous utilization of both optical and electrical properties can offer enhanced accuracy and detection versatility. Here, we demonstrate a dual-mode optoelectronic biosensor based on monolayer molybdenum disulfide (MoS₂) capable of producing simultaneous electrical and optical readouts of biomolecular signals. On a single platform, the biosensor exhibits a tunable photonic Fano-type optical resonance while also functioning as a field-effect transistor (FET) based on an optically transparent gate electrode. Furthermore, chemical vapor deposition grown MoS₂ provides a clean surface for direct immobilization of a water-soluble variant of the μ -opioid receptor (wsMOR), via a nickel ion-mediated linker chemistry. We utilize a synthetic opioid peptide to show the operation of the electronic and optical sensing modes. The responses of both modes exhibit a similar trend with dynamic ranges of four orders of magnitude and detection limits of <1 nM. Our work explores the potential of a versatile multimodal sensing platform enabled by monolayer MoS₂, since the integration of electrical and optical sensors on the same chip can offer flexibility in read-out and improve the accuracy in detection of low concentration targets.

Keywords: transition metal dichalcogenides (TMDCs), molybdenum disulfide (MoS₂), multimodal biosensor, photonic crystal, field-effect transistor, mu-opioid receptor

1. Introduction

Two-dimensional materials such as graphene and transition metal dichalcogenides (TMDCs) have been widely studied as promising functional materials in various sensing devices [1–5]. Among these, as a naturally available material, MoS₂ stands out as a versatile candidate due to its unique properties, namely the relatively high carrier mobility, suppressed off-state current due to a large electronic bandgap, and structural and chemical stability at ambient temperatures [6,7]. In light of these properties, great efforts have been devoted to developing MoS₂-based nanoelectronic biosensors for detecting a variety of biomolecular interactions, including DNA [8], prostate specific antigen (PSA) [2,3,9], and glucose [4]. Most of these biosensors operate in a single mode, particularly in a field-effect transistor (FET) configuration, utilizing the changes in channel conductivity and consequently the transduced electrical output as a result of binding or adsorption of target molecules. These binding events occurring near the surface also produce perturbations in the local dielectric permittivity enabling detection via electromagnetic means, which have been elusive. To this end, by coupling MoS₂ with a nanostructure exhibiting photonic resonances [10–12], one can realize a versatile multimodal sensor capable of interrogating biomolecular interactions with simultaneously transduced electrical and optical signals in one single device platform.

Multimodal biosensors can potentially improve the overall accuracy of detection by comparing results from different mechanisms, obtaining multiple characteristics of a biomolecular binding event, and providing flexibility for operation in different situations. Previous works have incorporated the two modes by fabricating plasmonic nanostructures on a TMDC surface [13], such that the device exhibits a photonic resonance that spectrally shifts upon attachment of target molecules [14,15]. The integration of graphene [14] or carbon nanotubes [15] into traditional devices have been demonstrated to enable parallel transduction of electronic, optical, and even mechanical signals. However, 2D materials in these devices merely act as a transparent substrate in the optical mode, playing no active role in optical transduction.

Here, we experimentally demonstrate a dual-mode biosensor as shown schematically in Figures 1a and b. By integrating MoS₂ monolayers on a dielectric photonic crystal (PhC) slab with an optically transparent gate electrode, MoS₂ takes an active part in both electrical and optical detection of DAMGO ([D-Ala², N-MePhe⁴, Gly-ol]-enkephalin), a synthetic opioid peptide and specific MOR agonist. We choose CVD-grown MoS₂ as the active material because the large (> 50 μm lateral size) and dense crystals make it suitable for large scale fabrication of devices [16,17]. In this device, monolayer MoS₂ plays critical roles as a functionalization

layer for nanophotonic sensing and FET channel for nanoelectronic sensing.

MoS₂ can be directly functionalized with a synthetic water soluble μ-opioid receptor, wsMOR [18], via Ni²⁺ linkers [19]. Monolayer MoS₂ exhibits a pristine surface with no dangling bonds, making direct functionalization difficult. Yet, proximity of binding events to the active surface is one of the factors that determines the sensitivity of a biosensor. Most MoS₂-based biosensors rely on a silanized top dielectric layer (e.g. APTES functionalization), such as Al₂O₃ and HfO₂ films [1,3,9], to bind protein probes. This limits the probe-target interactions to occur several nanometers away from the active MoS₂ surface, reducing the expected device sensitivity. The direct functionalization method used here utilizes Ni²⁺ ions adsorbed on the MoS₂ surface to bind to the histidine tag on the wsMOR protein [20]. This allows the receptor-target interactions to take place very close to the surface of monolayer MoS₂.

As shown in Figure 1c, CVD-grown MoS₂ flakes are placed over an array of nanoholes on a 100 nm thick suspended silicon nitride (SiN) membrane. Figure 1d shows a scanning electron micrograph (SEM) of a MoS₂-integrated square lattice hole array that makes up a PhC slab patterned lithographically between two electrodes. This PhC nanostructure exhibits optical Fano resonances which are characterized by their asymmetric line shape observable in the transmission spectra [21]. The spectral position of these Fano resonances are highly sensitive to small changes in the local permittivity in the surrounding medium. This enables sensitive optical detection of biomolecular interactions via refractive index modulation [22]. On this device, the accumulation of target molecules on monolayer MoS₂ induces an effective increase in the refractive index surrounding the PhC slab, causing the Fano resonance peak to redshift by Δλ. We employ this spectral shift in the optical transmission spectra to quantify the concentration of target molecules.

Simultaneously, bound molecules modify the in-plane conductivity via electrostatic gating effects, which can be detected as a change in the channel source-drain current (ΔI_{DS}). Figure 1c shows a mechanically transferred MoS₂ monolayer covering the drain and source electrodes, creating a semiconducting channel. To enable nanoelectronic sensing, 25 nm alumina (Al₂O₃) and 30 nm indium tin oxide (ITO) films are deposited on the backside of the SiN PhC slab as optically transparent dielectric and back gate electrode, respectively. These crucially enable the device to support two independent modes of sensing on the same chip. Since the PhC hole arrays are etched through the SiN slab, the underlying Al₂O₃ film maintains physical and electrical isolation between MoS₂ and the ITO back gate. As a result, the device exhibits field-effect characteristics while remaining optically transparent to allow transmission measurements.

2. Results and Discussions

Having described the physical mechanisms involved in our sensor, we illustrate in Figure 1b the dual-mode operation of the device. In the FET or nanoelectronic mode, two electrical sourcemeters (Keithley 2400) were connected, respectively, to the back-gate (V_G) and drain (V_D) electrodes with the source electrode acting as the common ground. We first verified that the presence of the nanoholes on the SiN membrane did not have a significant effect on the FET characteristics. To do this, we fabricated two devices, one with PhC hole arrays etched through the SiN_x membrane and one without. We then measured the gate-source current (I_{GS} , through ITO/Al₂O₃/SiN/MoS₂) as a function of V_G as shown in Figure 2a. The two devices displayed similar I_{GS} dependence on V_G , indicating that the hole arrays etched into the SiN dielectric layer did not inhibit the nanoelectric operation. The thin Al₂O₃ layer here is the key component in providing excellent electrical insulation between regions of MoS₂ suspended over nanoholes and bottom ITO back gate electrode. Without the Al₂O₃ layer, suspended monolayers can sag and come into contact with ITO under bias, which leads to electrical shorting through the structure. While this multilayer dielectric configuration provides electrical insulation on par with SiN, we still need to determine the range of voltages for which there is no dielectric breakdown for reliable operation of the dual-mode sensor. Experimental results in Figure 2a show that the leakage current I_{GS} increases exponentially when V_G is larger than 10 V indicating the onset of electrical breakdown. For $V_G > 20$ V, the capacitor structure can experience irreversible dielectric breakdown [23]. Therefore, for device characterization and sensing experiments, we limit the operation of the dual-mode sensor to gate voltages between $V_G = -2$ V and 5 V.

Next, we examined the transfer characteristics, I_{DS} - V_G and I_{DS} - V_D , of the MoS₂ FET. As shown in the I_{DS} - V_G curve in Figure 2b, we swept V_G from -2 V to 5 V at 1 V/s while keeping V_D constant, and measured I_{DS} across the electrodes. As expected, I_{DS} increased exponentially with V_G when the device is in the on-state and increasing V_D increases the channel current I_{DS} . In Figure 2c, we fixed V_G to obtain a relationship between I_{DS} and V_D . Device drain current I_{DS} increased linearly with V_D in the linear region and plateaued when the FET entered the saturation region at higher V_D bias. The I_G - V_{DS} and I_D - V_{DS} characteristics of our device is in good agreement with those of the previously reported MoS₂-based FETs [24–26].

In the optical or nanophotonic mode, the presence of hole arrays on a suspended SiN slab gives rise to Fano guided mode resonances that can be observed via transmission measurements. Upon illumination on a SiN PhC slab, a portion of the incident light is transmitted through the uniform slab, generating a slowly varying Fabry-Perot background (red line in Figure 2d). This hybridizes with the remaining

portion of light that excites the in-plane narrow-band guided resonance, produced by a periodic modulation in the dielectric constant over the square lattice hole array. As a result of this interference, we observe sharp Fano-type resonances in the transmission spectra characterized by their asymmetric spectral lineshapes [27]. The spectral positions of these Fano resonances can be tuned by varying the geometric parameters of the photonic crystal, such as the period and diameter of the holes.

We designed the photonic crystal to exhibit a Fano resonance at ~700 nm, spectrally away from the MoS₂ A-exciton absorption peak at 650 nm in order to minimize optical losses. Figure 2d shows the Finite Difference Time Domain (FDTD) method simulated transmission spectrum of a PhC consisting of a square array of holes with a radius of 120 nm and array period of 440 nm. The structure exhibits a Fano resonance at 690 nm with a theoretical quality (Q)-factor of about 60. The small spectral dip at ~650 nm corresponds to the A-exciton absorption peak of MoS₂ [28]. For transmission spectrum measurements, a collimated white light source was incident on the PhC and the transmitted light was collected through a microscope objective. Figure 2e shows the measured transmission spectra for the fabricated structure having the same PhC parameters, with and without the MoS₂ monolayer. The spectral peaks here are in good agreement with the FDTD simulations, with the A-exciton peak at 650 nm and the first and second Fano resonances at ~700 nm and 600 nm, respectively. Additional peaks in the measured spectra are attributed to nonuniformities introduced by the fabrication process.

To demonstrate the nanoelectric and nanophotonic sensing operation in this work, we utilized a histidine-tagged water soluble variant of the μ -opioid peptide receptor (MOR) and synthetic target opioid peptide analogue to enkephalin. The native MORs are a class of opioid receptors that possess a high affinity for endogenous ligands such as enkephalin and beta-endorphin [29]. Enkephalin is a neuropeptide which serves as an important signaling molecule in the brain [30]. Sensitive detection of opioid peptides is useful in drug discovery and development [19].

To achieve a clean surface for functionalization, a low temperature transfer method was employed as described in the methods section. This allows the PMMA layer to readily dissolve in acetone without leaving any residues on the MoS₂ surface. Atomic force microscope (AFM) images of MoS₂ surfaces before and after transfer in are shown Figure S1 (Supplementary Information). These confirm that the low temperature method yields a clean MoS₂ surface with very few residues. If a significant number of residues was present, functionalization would result in a lower density of immobilized probes available for detection, which would deteriorate the linear range, sensitivity, and limit of detection of the biosensor [31]. PMMA residues can also induce

unintentional doping effects and scattering loss in the nanoelectronic and nanophotonic operation, respectively, which would further degrade the detection sensitivity of the sensor [32].

Figure 3a illustrates the functionalization scheme for wsMOR on MoS₂ via Ni²⁺ linkers. To functionalize MoS₂, we first placed droplets of 5 mM NiCl₂ solution onto the device and let incubate for one hour. In this step, Ni²⁺ ions formed coordinate bonding with sulfur atoms on the surface of MoS₂. The sample was immersed in a water bath for 5 minutes followed by another water bath for 20 minutes to remove any excess Ni²⁺ ion. Next, droplets of 10 μM wsMOR in a buffer solution were placed on the device array for one hour, allowing the histidine tag to bind to Ni²⁺. Finally, the sample was left in a water bath for 5 minutes to wash off unbound receptors and blown dry.

After functionalization, the immobilization of wsMOR was verified via atomic force microscopy. An SEM of a functionalized MoS₂ monolayer is shown in Figure 3b. Figures 3c and 3d show the surface morphology of MoS₂ flakes before and after functionalization, respectively, indicating uniform protein coverage. The surface of transferred MoS₂ was uniform and relatively flat. The flake thickness was about ~1 nm (Figure 3e) which agrees well with the reported AFM measurements of CVD-grown monolayer MoS₂ [16,33,34]. After functionalization with Ni²⁺ and wsMOR protein, the thickness of MoS₂ flakes uniformly increased to ~3 nm (Figure 3f). This change is consistent with the 46 kDa size of wsMOR [20]. The surface of the substrate remained flat and featureless which indicates specific attachment of wsMOR on MoS₂ surface only. This mechanism of protein attachment on MoS₂ allows binding of target molecules to occur very close to the sensing surface.

As a result of wsMOR functionalization, the drain current of the device increased by ~8x indicating an increase in channel conductance (Figure 4a). The hysteretic behavior between the forward and backward voltage sweeps are attributed to adsorption of ambient molecules on the surface of MoS₂ and charge trapping between the monolayer and substrate [35]. We note that the functionalization process may either increase or decrease MoS₂ channel conductance depending on the quality and initial doping of the MoS₂ flakes. Meanwhile, the Fano resonance red-shifted by 3 nm due to the increased refractive index surrounding the MoS₂-integrated PhC slab (Figure 5a).

Dual-mode sensing experiments were carried out with target DAMGO synthetic opioid peptide diluted in deionized water. To test the sensor response, we exposed the device to target DAMGO in DI water at different concentrations, ranging from 0.1 nM to 10 μM. For each concentration, droplets of target DAMGO solution were placed on the device for 30 minutes. Then the device was rinsed in a DI water bath for 2 minutes and blown dry with a nitrogen gun. We note that

we do not expect the target DAMGO to bind to the Ni²⁺ linkers as they do not contain histidine residues (Ni²⁺ binding sites) as in the histidine-tagged wsMOR [36]. Therefore, a blocking agent was not necessary to deactivate the linker ions.

Current and transmission measurements were carried out sequentially at each concentration to obtain the dual-mode sensor responses. The dual-mode sensing responses as a function of DAMGO concentration are presented in Figures 4b and 5b, respectively, as a percent change in channel current, $\Delta I_{DS}/I_0$, and resonance peak shift, $\Delta\lambda$, with respect to responses of the functionalized device. For nanoelectronic sensing, we swept V_G from -2 V to 5 V at 1 V/s and a constant V_D of 0.1 V. This small value of V_D were used to keep the channel current small in order to avoid any heating effects. The current responses for different DAMGO concentrations were extracted at $V_G = 2.5$ V, half way between the turn on voltage and maximum voltage applied. For nanophotonic sensing, the measured transmission spectra were first fitted near the first Fano resonance. The peak positions were then obtained from the fit for each DAMGO concentration. Both curves were fitted by the Langmuir-Hill equation

$$\theta = A \frac{[C]^n}{(K_a)^n + [C]^n} + Z$$

where $\theta = \Delta I_{DS}/I_{DS,0}$ or $\Delta\lambda$ is the sensor response representing the fraction of wsMOR that is bound to DAMGO, $[C]$ is the DAMGO concentration, K_a is the DAMGO concentration that produces half of the maximum response, A is the maximum response, Z is the offset response, and n is the Hill coefficient. The dissociation constant describing binding affinity is described as $K_d = (K_a)^n$.

Responses in both modes exhibit similar trend as they scale with the fraction of conjugated wsMOR on MoS surface. The electrical response in Figure 4b is fitted with $K_a = 1.2$ nM, $A = 145$ and $n = 0.87$, and the optical response in Figure 5b with $K_a = 1.0$ nM, $A = 1.86$, and $n = 0.91$. Figures 4c and 5c show normalized response of three devices. We extracted averaged fitting parameters to be $K_a = 1.5 \pm 0.4$ nM and $n = 0.87 \pm 0.01$ for the nanoelectronic response, and $K_a = 1.2 \pm 0.2$ nM and $n = 0.89 \pm 0.02$ for the nanophotonic response. These yield the dissociation constants, K_d , for DAMGO-wsMOR interaction of 3.6 ± 0.5 nM and 2.5 ± 0.8 nM as obtained via electronic and optical modes, respectively. These values are also in good agreement with previously reported K_d of MOR for DAMGO [37,38]. Despite using different transduction mechanisms, we found the two K_d values to be remarkably close. This evidently demonstrates the independent operation of the two systems, allowing us to consistently probe biomolecular binding events via two distinct physical mechanisms on the same device.

For both modes, the limit of detection is about 0.1 nM as limited by electrical noise and the spectral resolution of the instrument. Since the sensor responses both scale with surface coverage of the target peptide, the linear range of the sensor is approximately 0.1–10 nM for both modes of operations. The

sensitivity of the sensor is extracted from the slope of the fit in the linear range, yielding the sensitivity as a percent change of drain current per unit target concentration of 10.5% /nM, and spectral displacement per unit concentration of 0.15 nm/nM. We also note that the sensing area and photonic crystal design on this device can be independently optimized further to improve the sensitivity while maintaining dual-mode operation. The combination of two sensing modes can further improve the accuracy of detection by reducing the margin of errors in the individual sensor responses, and provide great flexibility in modes of operation.

3. Conclusion

In summary, we demonstrate here a dual-mode optoelectronic biosensor based on CVD-grown monolayer MoS₂ as an active multifunctional element. The device is composed of a MoS₂-based FET fabricated on a SiN PhC slab exhibiting an optical Fano resonance. ITO back gate electrode and Al₂O₃ dielectric layer keep the device transparent for transmission measurements, and prevent electrical shorting through the nanoholes. These allow the device to support two modes of sensing simultaneously. The monolayer MoS₂ acts as a protein immobilization surface, active functionalization layer for nanophotonic refractive index sensing via Fano resonance tuning, and FET channel for nanoelectronic sensing. With our present design, the percent yield of working devices is about 70%, due to the flakes having to both bridge two electrodes and cover a photonic crystal hole array. We note that a different device layout could increase the yield further. Our previous work demonstrated a large scale production of similar FET biosensors using CVD-grown MoS₂, achieving a yield of over 90% [20]. Therefore our proof-of-concept demonstration of the dual mode sensor can readily be extended to large area fabrication with reasonably high yield in the future. To demonstrate dual-mode operation, the sensor was functionalized with wsMOR, and used to determine the affinity of DAMGO as a function of concentration via electrical and optical measurements. The sensor responses were read out in terms of percent change in drain current I_{DS} and wavelength shift for the Fano resonance in the nanoelectronic and nanophotonic modes, respectively. Both responses agreed well with the Langmuir-Hill model for ligand-receptor binding and exhibited binding constants within the same range as those previously reported in literature using radio-active materials. This approach not only avoids using radio-active materials, but also combining two readouts can increase the accuracy of results by reducing uncertainty that arises from each individual mode, such as calibration and systematic errors. The design can potentially be optimized to provide extensive measurements of biomolecular processes, improved dynamic sensing range and sensitivity and flexibility in mode of operation. Our work shows that the

unique physical, chemical, electronic and optical properties of monolayer MoS₂ can open up new opportunities for the design of multimodal biosensors.

4. Methods

4.1 Device Fabrication

An array of devices were fabricated on a freestanding 100 nm-thick silicon nitride membrane (Norcada), which provided a transparent substrate for transmission measurements. First, 20 nm Al₂O₃ was grown on the backside of the substrate by Atomic Layer Deposition (ALD) at 200 °C. This was followed by deposition of 30 nm ITO by RF sputtering. Next, we used photolithography to pattern electrodes on the front side of the substrate; 5 nm Cr and 50 nm Au were deposited via Electron-Beam Evaporation and lifted off to form electrodes. Hole arrays were patterned in between electrodes on the SiN membrane via Electron-Beam Lithography. The holes were formed by dry etching through the SiN slab via Reactive Ion Etching (RIE) with O₂ and CHF₃ gases with Al₂O₃ layer on the back side acting as an etch stop layer.

4.2 CVD MoS₂ Transfer

After the electrodes and hole arrays were fabricated, flakes of CVD-grown MoS₂ were transferred onto the devices by wet transfer. We first spin-coated PMMA (500 nm – 1 um thick) on as-grown MoS₂ flakes on a SiO₂/Si substrate. Next, the sample was left floating in a bath of 0.1M NaOH to etch SiO₂, releasing the PMMA film from the underlying substrate. The floating PMMA film was carefully scooped up with a glass slide and released in three sequential DI water baths to wash off any NaOH. The PMMA film was then scooped up from the last water bath with the patterned SiN membrane chip, and the sample was blown dry with a N₂ gun until the water droplets visually disappeared. Then the sample was left to further dry in a dessicator for at least two hours. To remove PMMA, the sample was submerged in an acetone bath for 30 minutes, followed by a second acetone bath for 1 hour. Finally, the sample was placed in an IPA bath for 10 minutes to remove acetone and blown dry. The finished device underwent a current annealing process prior to measurements. With constant V_G=10 V, V_D was slowly increased from 0.1 V to 3 V and left for 2 minutes. Prior to sensing experiments, transferred MoS₂ flakes on the device were functionalized with wsMOR as described in main text.

Acknowledgements

This work was supported by NSF under the NSF 2-DARE Program (EFMA-1542879), and NIH R01 (1R01GM111421, PI, RL).

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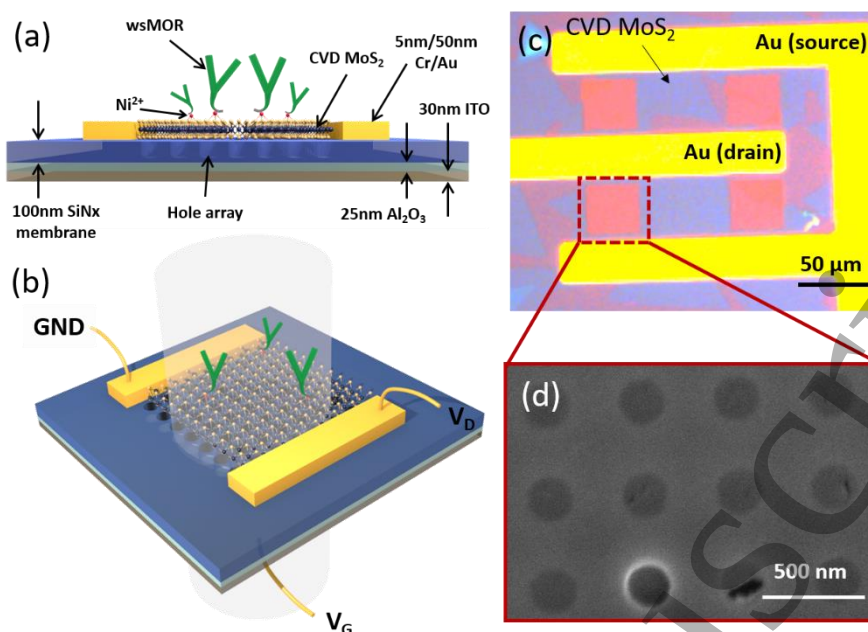


Figure 1. Device structure and sensing scheme (a) The biosensor was fabricated on a freestanding 100 nm-thick SiN membrane supported by a Si frame. On the backside of the membrane are 25 nm Al₂O₃ and 30 nm ITO. Photonic crystal slabs were fabricated directly on the SiN membrane between two Cr/Au electrodes. Large CVD MoS₂ flakes were transferred over the two electrodes and photonic crystals to create an FET and optical modulator on the same platform. (b) Sensing scheme of the dual-mode sensor with V_G and V_D connected to the gate and drain electrode, respectively, with the source electrode as a common ground. The nanohole arrays have a period of 440 nm and hole radius of 120 nm, giving rise to a Fano resonance at ~700 nm for refractive index sensing. (c) Microscope image of a device. Purple-ish blue regions are the overlaying CVD MoS₂. (d) SEM of a PhC hole array with the superimposing monolayer CVD MoS₂.

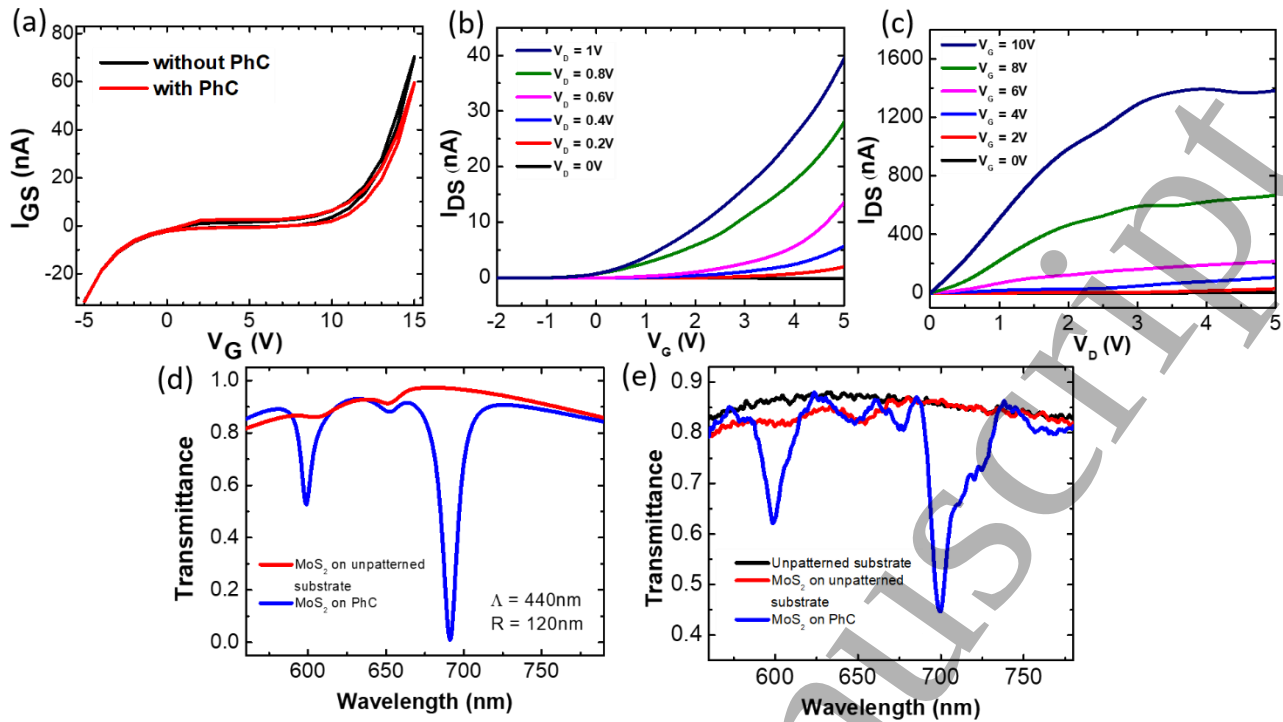


Figure 2. Optoelectronic device characterization. (a) Comparison between I_{GS} - V_G curves for the device with and without PhC hole arrays. (b) I_{GS} - V_D and (c) I_{DS} - V_D characteristics for the MoS₂-based FET. (d) Simulated FDTD transmission spectra of MoS₂ on an unpatterned (red) and patterned (blue) substrate. The substrate is a 100nm-thick suspended SiN slab with 25 nm Al₂O₃ and 30 nm ITO layers on the backside. The red curve shows a slowly varying Fabry-Perot resonance due to the slab substrate. Small peaks at 650 nm and 610 nm are due to the absorption of A- and B-excitons, respectively. Patterned substrate consists of a rectangular array of nanoholes etched only through the SiN slab. The blue curve shows peaks at 690 nm and 600 nm corresponding to the first and second order Fano resonances, respectively. These resonances arise from the hybridization of 2nd order grating scattering from the nanohole array and the Fabry-Perot transmission background. (f) Measured transmission spectra for the fabricated structure. Additional peaks are attributed to nonuniformities in the fabrication process.

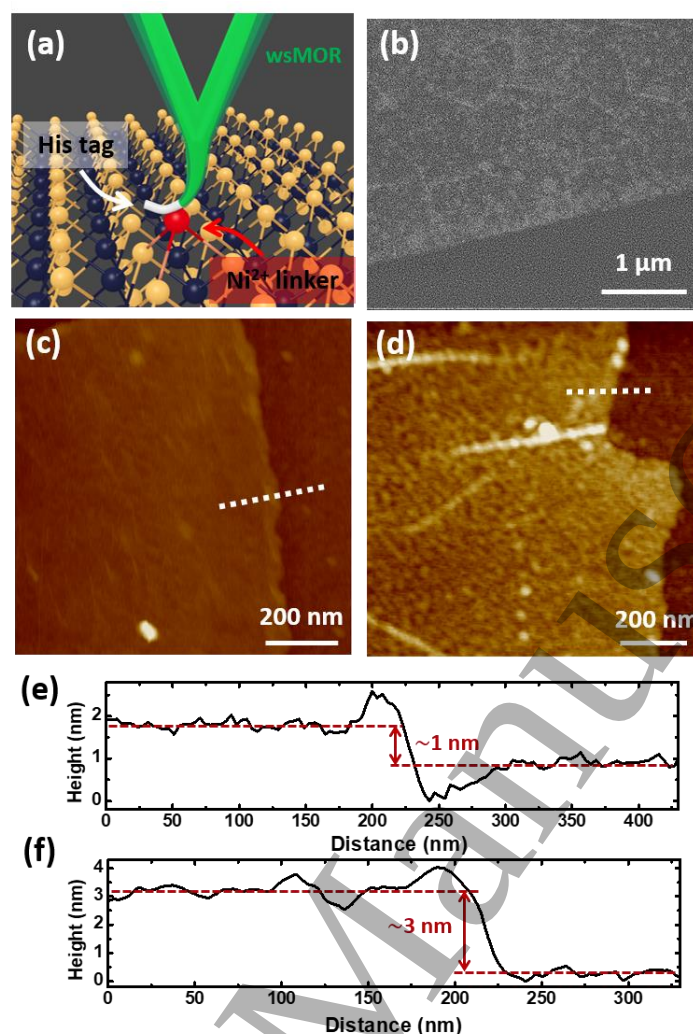


Figure 3. MoS₂ functionalization. (a) Schematic for the linker chemistry for direct functionalization. The red atom represents Ni²⁺ that is bound to the sulfur atoms on MoS₂ via coordinate bonds. The white region on wsMOR represents the histidine tag that binds preferentially to the Ni²⁺ ions. (b) SEM of wsMOR-functionalized MoS₂. (c) AFM topographic image of unfunctionalized transferred CVD MoS₂ on SiO₂, (d) wsMOR-functionalized MoS₂ on SiO₂. (e,f) Height profiles of the line scan shown in (c) and (d), respectively.

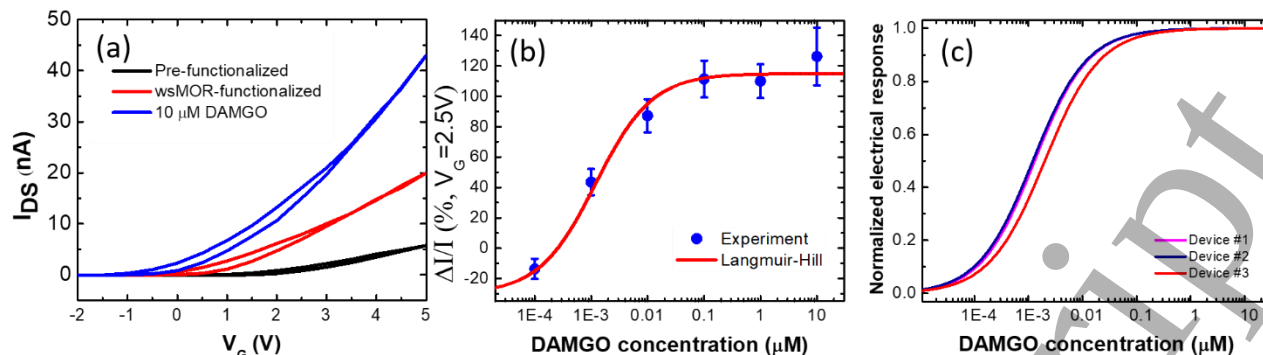


Figure 4. Nanoelectronic sensing. (a) I_{DS} - V_G characteristics of the sensor pre-functionalization (black), after functionalization with wsMOR (red), and after exposure to 10 μ M DAMGO (blue). $V_D = 0.1$ V. (b) Electronic response reported as percentage change in I_{DS} with respect to the I_{DS} of wsMOR-functionalized device, extracted from I_{DS} - V_G curve at $V_G=2.5$ V. Error bars represent the standard deviation of the mean between three repeated measurements. The experimental data was fitted by the Langmuir-Hill isotherm to obtain the sensor calibration curve for the nanoelectronic mode (red). (c) Normalized electronic response of three different devices.

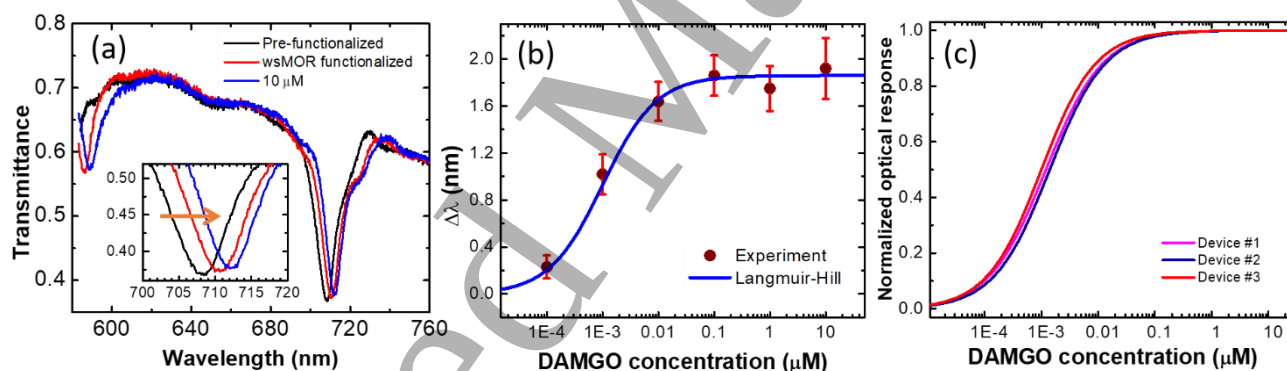


Figure 5. Nanophotonic sensing. (a) Transmission spectra of the sensor pre-functionalization (black), after functionalization with wsMOR (red), and after exposure to 10 μ M DAMGO (blue). (b) Optical response reported as resonance wavelength shift with respect to the resonance peak position of wsMOR-functionalized device. Error bars represent the standard deviation of the mean between three repeated measurements. The experimental data was fitted by the Langmuir-Hill isotherm to create the sensor calibration curve for the nanophotonic mode (blue). (c) Normalized optical response of three devices.