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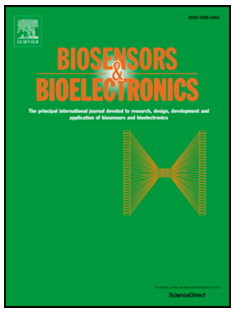

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CRediT authorship contribution statement

Pengfei Song: Conceptualization, Writing-Original Draft, Writing-Reviewing and Editing. **Hao Fu:** Validation. **Yongjie Wang:** Validation. **Cheng Chen:** Investigation. **Pengfei Ou:** Investigation. **Roksana Tonny Rashid:** Validation. **Sixuan Duan:** Writing-Reviewing and Editing. **Jun Song:** Conceptualization. **Zetian Mi:** Conceptualization. **Xinyu Liu:** Conceptualization, Writing-Reviewing and Editing.

A Microfluidic Field-Effect Transistor Biosensor with Rolled-Up Indium Nitride Microtubes

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Abstract

Field-effect-transistor (FET) biosensors capable of rapidly detecting disease-relevant biomarkers have long been considered as a promising tool for point-of-care (POC) diagnosis. Rolled-up nanotechnology, as a batch fabrication strategy for generating three-dimensional (3D) microtubes, has been demonstrated to possess unique advantages for constructing FET biosensors. In this paper, we report a new approach combining the two fascinating technologies, the FET biosensor and the rolled-up microtube, to develop the microfluidic diagnostic biosensor. We integrated, an excellent biosensing III-

nitride material—indium nitride (InN)—into a rolled-up microtube and used it as the FET channel. The InN possesses strong, intrinsic, and stable electron accumulation (~10^{13} cm^{-2}) on its surface, thereby providing a high device sensitivity. Multiple rolled-up InN microtube FET biosensors, which were fabricated on the same substrate, was also integrated with a microfluidic channel for convenient fluids handling and shared the same external electrode (inserted into the microchannel outlet) for gating voltage modulation. Using human immunodeficiency virus (HIV) antibody as a model disease marker, we characterized the analytical performance of the developed biosensor and achieved a limit of detection (LOD) of 2.5 pM for serum samples spiked with HIV gp41 antibodies. The rolled-up InN microtube FET biosensor represents a new type of III-nitride-based FET biosensor and holds significant potential for practical POC diagnosis.

Introduction

Miniaturized and sensitive biosensors capable of rapidly detecting disease-relevant biomarkers at the point-of-care (POC) have enormous impact on healthcare by enabling early-stage disease diagnosis and timely treatment (Yager et al. 2008). Conventional diagnostic tests are performed in clinical laboratories, and may take long turnaround time (hours and even days) for potential patients to get the results. This may cause delayed treatment and impose unnecessary anxieties/stresses on the patients. For patients carrying infectious diseases, the long waiting time before determinative diagnosis may lead to further spreading of the diseases due to the patients' unawareness of their conditions (Gubbins et al. 2014; Qin et al. 2020). POC devices offer an excellent solution for providing accurate, quality-assured tests near patients in a rapid fashion (e.g., in minutes), and can expedite diagnosis and treatment initiation in under-served regions/communities and/or in resource-poor settings (e.g., developing countries) (Fu et al. 2019; Pant Pai 2007; Vickerman et al. 2006; Zhao and Liu 2016).

There has been a significant technological trend of developing POC devices to meet the urgent demands for timely and sensitive diagnostic services. With many apparent advantages suitable for POC applications, FET biosensors have emerged as an enabling platform technology for developing sensitive and rapid POC tests (Kwon et al. 2012; Shoorideh and Chui 2014). FET biosensors usually feature fast response, label-free assay, high sensitivity (enabled by the high sensitivity of the FET channel material to surface binding of molecules), good compatibility with batch fabrication, and thus low unit cost (Cui et al. 2001; Kauffman and Star 2008). The miniature nature of FET biosensors also permit their easy integration into microfluidic devices to gain additional advantages such as multiplexed testing (detection of multiple markers) and less sample/reagent consumptions (Ang et al. 2011; Xu et al. 2014).

Rolled-up nanotechnology, as a novel strategy that combines bottom-up thin-film growth/deposition and top-down lithography for fabricating three-dimensional (3D) tubular micro/nano-structures, has also shown unique advantages for developing microfluidic biosensors. First, the rolled-up microtubular structure has a small footprint and can be batch-fabricated and integrated into microfluidic channels for convenient fluids handling and multiplexed detection. The rolled-up microtube itself also naturally forms a microfluidic channel that can transport fluids for biosensing. The first rolled-up microtube fluidic sensor was reported in 2008 (Bernardi et al. 2008). After that, rolled-up microtube has been used to develop a variety of biosensors based on different sensing mechanisms, including optical ring resonator (Huang et al. 2010; Li et al. 2019; Song et al. 2018; Wang et al. 2019; Zhao et al. 2012), Raman spectroscopy (Yin et al. 2012; Zhang et al. 2015), electrical sensing (Egunov et al. 2021; Grimm et al. 2013), and magnetic sensing (Karnaushenko et al. 2018; Mönch et al. 2011), and self-propelled micromotor-based sensing (Han et al. 2016), to name a few. Second, the exceptional fabrication flexibility of rolled-up nanotechnology allows microtubes to be made from a variety of functional materials including semiconductor thin-films and thus serve as FET channel structures,

promising enhanced performance from the semiconductor microtube-based FET biosensors (Li and Mi 2009; Li 2011; Mei et al. 2008). Last, using a tubular structure as the FET channel has been demonstrated to enhance the sensitivity of the FET biosensor, by reducing the Debye screening effect (Shoorideh and Chui 2014). Because of the Debye screening existing on its channel surface, an FET biosensor is only responsive to the charged molecules within a certain length (the Debye length), and the device sensitivity is thereby severely limited (Stern et al. 2007).

This work aims to take full advantage of this merit to fabricate rolled-up indium nitride (InN) microtubes and use it to construct a new type of FET biosensor. Although various materials including 2D materials (Cui et al. 2018; Xu et al. 2017; Zhou et al. 2019), III-V group materials (Dastjerdi et al. 2015; Mei et al. 2009; Yu et al. 2015), and polymers (Hu et al. 2018; Jager et al. 2000; Zakharchenko et al. 2011) have been used to construct rolled-up microtubes, the semiconductive InN material has not been explored. InN nanostructures possess the highest intrinsic surface electron accumulation (~10^{13} cm^{-2}) among all III-nitride materials, and this inherent property of InN cannot be easily altered physically or chemically during microfabrication (Lu et al. 2004; Naoi et al. 2007). The high surface electron accumulation makes it highly sensitive to the binding of charged biomolecules, and thus ideal for serving as the channel structure of an FET biosensor (Chen et al. 2006; Lu et al. 2003; Mahboob et al. 2004). There has been only one attempt to developing microtube-based FET chemical sensors (Grimm et al. 2013). Grimm *et al.* developed the first rolled-up microtube FET (RUM-FET) biosensors using InGaAs/GaAs bilayers. This work only demonstrated the sensing capability of the RUM-FET using different solvents, and the detection of real disease markers was not demonstrated. In addition, the semiconductor materials used in this design, InGaAs and GaAs, have lower surface electron accumulation densities than InN.

In this paper, we report a new type of FET biosensor integrating high-performance InN tubular microstructures, and demonstrate label-free detection of human immunodeficiency virus (HIV)

antibodies. The InN thin films were deposited, through facile DC reactive sputtering, on a bilayer of strain-engineered SiO_x/SiN_x (bottom/top) nanomembranes, which served as the rolling vehicle to form SiO_x/SiN_x/InN microtubes. The material properties of the deposited InN were characterized using X-ray photoelectron spectroscopy (XPS) and scanning electron microscopy (SEM). The InN microtube is functionalized with capture proteins specific to the HIV antibodies using convenient (3-Aminopropyl) trimethoxysilane as the chemical linker, and the surface chemistries were characterized by XPS (Chen et al. 2006). Using the InN RUM-FET biosensor, we performed label-free detection of HIV gp41 antibodies, and achieve a limit of detection of 0.1 ng mL^{-1}.

Materials and methods

Materials and reagents

All the chemicals used in the experiments including (3-Aminopropyl)-trimethoxysilane (APTMS) (97%), 2.5% (v/v) glutaraldehyde (GA), 1 × phosphate buffered saline (PBS), 0.1% bovine

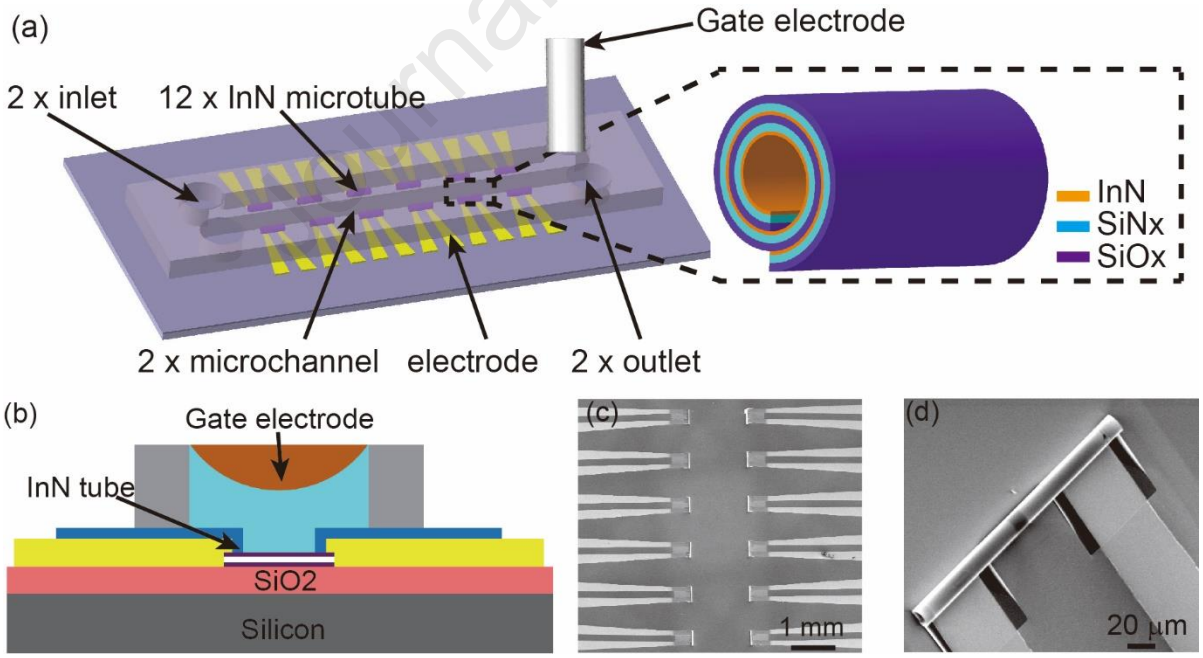

Figure 1. *(a) Schematics of InN RUM-FET biosensors enclosed in a microfluidic channel. The channel width is 400 μm and length is 6.5 mm, which is large enough to accommodating the 12 tube devices. (b) Schematic of the*

experiment condition with inserted electrode. (c) SEM photograph of an array of fabricated InN microtubes. (d) A high-magnification SEM photograph of an InN microtube, with zoomed-in view of the channel area.

serum albumin (BSA), were purchased from Sigma-Aldrich (Oakville, Canada), unless otherwise indicated. HIV pg41 antibody was purchased from ViroGen.

Design of rolled-up InN microtubes

As shown in Figure 1(a), the rolled-up microtubes were formed from three layers of SiO_x/SiN_x/InN nanomembranes (bottom to top). The bottom two layers of SiO_x/SiN_x, which are much thicker than the top InN layer (15 nm/50 nm vs. 5 nm), were deposited with controlled built-in stresses using plasma enhanced chemical vapor deposition (PECVD) and served as the driver for microtube self-rolling upon release from the substrate. The thicknesses of the bottom two layers heavily affect the yield of the rolling-up process, as the different thickness combinations lead to different strain gradient in the bottom two layers. We experimentally demonstrated that the combination of 15 nm SiO_x and 50 nm SiN_x yielded the best fabrication yield (Song et al. 2018). The InN film was set to be only 5 nm thick because of two reasons. First, a thick InN layer could jeopardize the microtube self-rolling process by affecting the strain gradient, preventing the SiO_x/SiN_x/InN nanomembranes from completely rolling up due to a large external strain gradient posed by the thick InN layer. Second, a thick sputtering-deposited InN film could be difficult to be patterned through the lift-off technique (see the "Fabrication of the rolled-up InN microtubes" section), as the edge of the sacrificial photoresist under the InN film will be also coated by the thick InN film. Once a SiO_x/SiN_x/InN microtube is rolled up, the thin layer of InN will be exposed on the inner wall of the microtube (Figure 1(a)), serving as the biosensing surface. The length of the microtube is 350 μm with a diameter of ~20 μm, and the microtubes in the same row are 650 μm away from each other. The FET channel length (the gap between the two electrodes) was designed to be short (20 μm) to enhance the FET sensing performance. The major design parameters of the FET were determined by considering the

microfabrication resolution, the arrangement of microtube arrays on the silicon wafer, and the general FET design rules for increasing the biosensing performance. For instance, reducing the FET channel length could improve the device sensitivity but may increasing the fabrication difficulty and thus sacrifise the fabrication yield (Fathil et al. 2018)."

The InN inner wall of the microtube forms an ideal concave surface contacting with the analyst solution, and this configuration has been theoretically determined to reduce the electrostatic screening on the biosensing surface and thus prove the device's sensitivity of charge detection in electrolytic environments (Shoorideh and Chui 2014). The micrometer size and batch fabrication process of the rolled-up InN microtube also allow the integration of multiple InN RUM-FETs inside a microfluidic channel to facilitate the solution delivery and reduce the sample/reagent consumptions. As shown in Figure 1(c), we batch-fabricated two rows of InN RUM-FET biosensors (six for each) on the same silicon (Si) substrate, and enclosed each row of the biosensor in a separate PDMS microfluidic channel. When running a test, the sample solution was delivered into each microchannel via its inlet, and a miniaturized Ag/AgCl reference electrode (ET073-1, eDAQ) was inserted into the microchannel outlet to modulate the FETs gating voltage (Fig. 1(d)). A previous study has shown that the use of a miniaturized reference electrode by multiple FETs does not cause performance inconsistency of the FETs (Xu et al. 2014). The diameter of the gate electrode is 2 mm, which is slightly larger than that of the outlet diameter (1.5 mm) of the microfluidic device. Therefore, once inserted into the device outlet, the electrode formed a stable mechanical and electrical connection.

Fabrication of rolled-up InN microtubes

The rolled-up microtubes were fabricated through controlled releasing of strain-engineered SiO_x/SiN_x/InN thin-films. This fabrication process was adapted from a previous protocol we reported for fabricating SiO_x/SiN_x microtubes (Song et al. 2018). As shown in Figure 2, the process started from

electron-beam deposition (BJD 1800, Temescal) of an aluminum sacrificial layer (50 nm) on a cleaned Si wafer (Figure 2a), followed by PECVD deposition (Oxford Plasma 100) of SiO_x/SiN_x (bottom/top) thin films (Figure 2b). The PECVD parameters of 8.5 sccm SiH_4 with 710 sccm N_2O, and 10 sccm SiH_4 with 10 sccm NH_3 were experimentally determined to produce suitable film stresses for SiO_x (–831.5±21.4 MPa, n = 5) and SiN_x (221.5±12.1 MPa, n = 5), respectively. The deposition temperature, power, and working frequency were 300 °C, 70 W, and 13.56 MHz for both layers. The SiO_x/SiN_x bilayer acts as the rolling vehicle to roll-up the subsequently deposited InN thin film. Then, the SiO_x/SiN_x bilayer was patterned into a U-shape through standard photolithography (photoresist: S1813, MicroChem) and wet etching by 10:1 diluted hydrofluoric acid.

Once the U-shaped SiO_x/SiN_x was formed, InN thin film was deposited and patterned into the same U-shape, and was precisely aligned on the top of the SiN_x/SiO_x bilayers (Figure 2c). The 5 nm InN thin film was deposited by DC reactive sputtering (E14, Denton). Briefly, the pure indium target (99.999%) was used under the argon and nitrogen gas environment (Ar 20 sccm, and N_2 100 sccm). The DC current was set to 0.22 A and substrate temperature was maintained 45 °C. After the growth time of 120 second, one lift-off process was performed using the S1813 photoresist (MICROPOSIT) and 1165 solution (65 °C for at least 30 minutes). Two gold electrodes were then fabricated on top of the InN layer by performing another lift-off of the e-beam (BJD 1800, Temescal) deposited biolayer of titanium/gold (5 nm/15 nm), leaving a 20 μm gap in between as the biosensing area of the FET channel. The titanium/gold electrode is used to form the Ohimc contact with the InN layer according to the previous reports (Chang et al. 2010; Oseki et al. 2014). The atomic layer deposition (ALD; GEMStar-8™) is finally used to insulate the electrode to avoid any exposure to the analyte solution. Another photolithography is performed for patterning a rectangular photoresist (S1813, MICROPOSIT) structure covering the legs of the U-shapes, which acted as the anchor of the SiO_x/SiN_x/InN nanomembranes during the releasing/rolling-up process. The releasing process was

performed using the aluminum etchant type A (Transene Inc.) under elevated temperature (80 °C) to allow the SiO_x/SiN_x/InN nanomembranes to self-rolling upon the removal of the aluminum layer. The sample was then immersed in deionized (DI) wafer for stopping the etching. A critical point dryer (CPD, Tousimis) was finally used to dry the rolled-up microtubes; otherwise, the large surface tension during the air-environment drying process could lead to the microtube collapse.

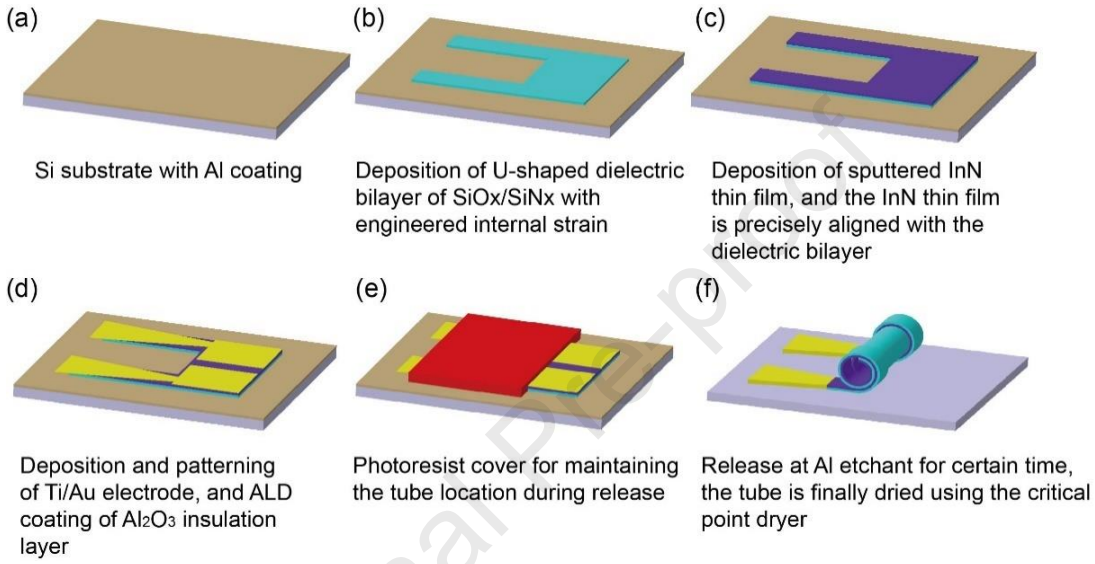

Figure 2. Fabrication process of the rolled-up InN microtube device*. (a) The pre-cleaned silicon wafer was coated with 50 nm Al sacrificial layer. (b) The strain-engineered SiO_x/SiN_x bilayer was deposited by PECVD and patterned by wet etching. (c) The InN thin film was then deposited by sputtering and patterned by lift-off. (d) Another lift-off was performed for pattering the e-beam deposited Ti/Au electrode. The ALD coating of Al_2O_3 was also performed to cover the electrode surface for preventing the direct solution contact with the electrode. (e) Photoresist cover was used for maintaining the tube location during the release. (f) The tube was finally released from the substrate using the Al etchant and dried at CPD before use.*

InN material characterization

We utilized a variety of tools to characterize the synthesized InN thin film, including the X-Ray photoelectron spectroscopy (XPS, Thermo-Scientific *K*-Alpha), and scanning electron microscopy (F50, FEI). The high-resolution XPS spectra with an energy resolution smaller than 0.2 eV were

collected using Al-Kα ($h\nu = 1486.6$ eV) as the X-Ray beam source. All XPS spectra were corrected and aligned using C *1s* (284.8 eV) as the reference.

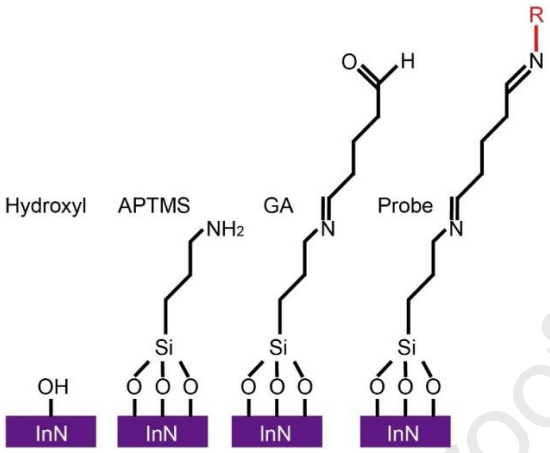

Figure 3. *Schematic diagram of the surface chemistry route for biofunctionalization InN thin film. The probe protein molecule in the final step is denoted as 'R'.*

InN surface biofunctionalization and device packaging

We employed an organosilane-based surface functionalization process (Figure 3) to covalently bind the capture proteins on the InN surface, which was adapted from a previously established method . Briefly, the rolled-up InN microtube was first treated and activated by oxygen plasma (DSB6000, NanoPlas) for 10 minutes at 10 watts power and 0.6 mTorr pressure, during which the substrate was maintained at 25 °C. The plasma activation process led to a dense layer of hydroxyl groups (–OH) on the InN surface and the Si substrate. We then immediately bonded the pre-prepared DPMS layer, which includes two microchannels fabricated through soft lithography, on the Si substrate to cover each row of RUM-FETs with one microchannel. The (3-aminopropyl)-trimethoxysilane (APTMS, 97%) solution was then infused into the microchannel to bind with the InN surface through Si-O bonds (Figure 3). After 30-minute reaction, we thoroughly washed the rolled-up microtubes with ethanol to remove extra APTMS. The 2.5% (v/v) glutaraldehyde (GA) solution (10 mM) was then infused into the

microchannel for reacting with the amine ($–NH_2$) groups of APTMS, and providing aldehyde (CHO-) groups for subsequent capture probe immobilization. The reaction of APTMS and GA was also incubated for 30 minutes. After the GA treatment, the rolled-up microtubes were thoroughly washed with DI water, and then treated with 10 μg mL^{-1} HIV gp41 antibody solution for 30 minutes at room temperature. In the end, 1× PBS solution was used to wash off the unbound antibodies. To avoid non-specific binding, we also used 0.1% bovine serum albumin (BSA) solution to block the InN surface through 10-minute incubation followed by thorough washing by 1× PBS. All the aforementioned surface reaction steps were carried out at room temperature (21 °C).

Device characterization

The electrical characteristics of the InN RUM-FET were measured by a precision dual-channel source meter (2602, Keithley). For measurements of the transport characteristics of the RUM-FET, the source-to-drain voltage (V_d) was set to 0.1 V and the solution gate voltage V_g was scanned from −0.5 V to 0.5 V. The sweep step of V_g was 100 mV, and for each step the applied V_g pulse was maintained for 2 s to stabilize the I_{ds} to ensure the reliability of the transport curve.

Results and discussion

Characterization of the InN thin film

Though molecular beam epitaxy (MBE) is widely used to produce high-quality InN thin films for device applications, the high temperature (550 °C) involved in the MBE growth could induce annealing effect on the strain-engineered SiO_x/SiN_x layers and thus release the built-in stresses of the nanomembranes. Therefore, it is not feasible to the MBE method to integrate InN into the rolled-up microtubes. We adopted the DC reactive sputtering for growth of the InN thin film on the SiO_x/SiN_x bilayer, which is a low-temperature process and compatible with the fabrication of rolled-up microtubes (Lindgren et al. 2006). To verify the quality of the sputtered InN film, we first examined the

chemical composition of the InN film by XPS. The XPS spectra, as shown in Figure 4a and 4b, were collected from the InN thin film deposited on a SiN_x layer (to mimic the SiO_x/SiN_x bilayer). The In *3d* spectrum

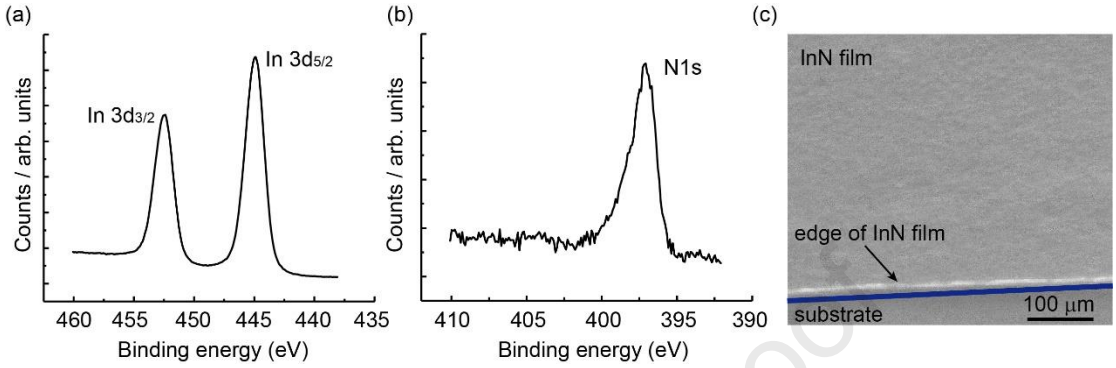

Figure 4. Characterizations of the sputtering synthesized InN thin film. (a) In 3d core-level XPS spectra, (b) S 2p core-level XPS spectra, and (c) SEM image of synthesized InN thin film.

has two peaks at 443.8 eV and 451.4 eV (Figure 4a), which are attributed to the doublet of In $3d_{5/2}$ and In $3d_{3/2}$, respectively. The N *1s* spectrum has one peak at 396.6 eV (Figure 4b), corresponding to the nitrogen bonded with indium. The measured XPS spectra are in good agreement with previously reported data (Lindgren et al. 2002), confirming the successful growth of InN. Figure 4c shows the SEM photograph of the 5 nm InN thin film deposited on the top of the SiN_x layer. It can be seen that the InN film shows continuous and highly uniform surface morphology.

Verification of the InN surface biofunctionalization

After successfully growth of the InN films, we experimentally verified the surface functionalization of the InN film with (3-aminopropyl)-trimethoxysilane (APTMS), which serves as the chemical linker for immobilizing capture proteins specific to the target protein marker (Chen et al. 2006; Li and Liu 2016). XPS was used for probing the surface chemical change caused by surface chemical functionalization. For the InN film functionalized with APTMS, the bonding of APTMS molecules onto the plasma-treated InN surface was verified by observation of the XPS N *1s* spectra

before and after the APTMS treatment (Chen et al. 2006). Figure 5 shows the N *1s* core-level spectra unequivocally. Two new peaks at 398.4 eV and 400.6 eV were observed after the APTMS treatment (Figure 5b). These two peaks correspond to the NH_2 and NH^{3+}, respectively, which are due to the immobilization of APTMS on

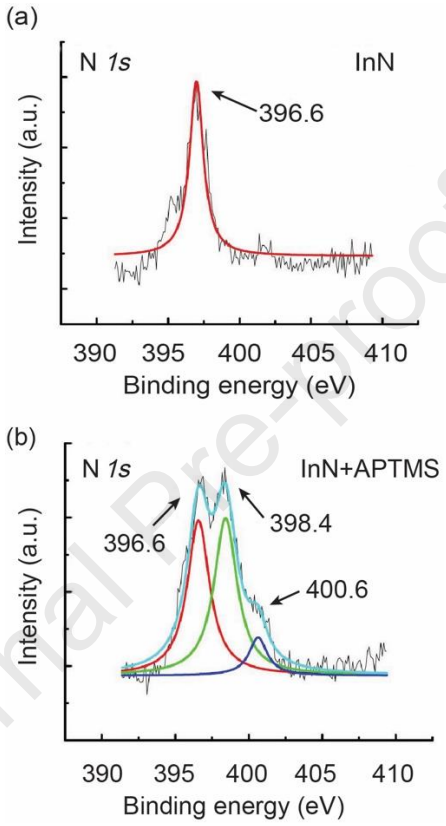

Figure 5. *N 1s core-level XPS spectra of (a) pristine and (b) APTMS-coupled InN thin films. The XPS spectra (black lines) are fitted to show the individual peaks (color lines).*

the InN surface. For the pristine InN sample, only 396.6 eV peak was detected (Figure 5a). These observations well agree with the previously reported results (Chen et al. 2006), suggesting that our APTMS-based surface functionalization is effective.

Electrical characterization of the InN RUM-FET

We first characterized the electrical properties of the fabricated InN RUM-FET. The device's conducting and transporting properties after biofunctionalization were measured, as shown in Figure 6(a) (b). It can be seen from Figure 6a that source-to-drain current increases linearly in the measured voltage range, indicating the formation of Ohmic contacts. After biofunctionalization, the conductance of the FET channel decreased from its intrinsic value of 25 mS to 2.4 mS. The decrease of the channel conductance after surface biofunctionalization has been commonly observed in other similar studies (Kwon et al. 2012; Mao et al. 2013; Mao et al. 2011). The main reason is that after adding the biomolecules to the conducting FET channel, the local geometric deformation and the number of scattering centers across the InN channel was introduced, thereby decreasing the FET channel conductance. We also performed the solution-gate transistor measurement of the RUM-FET in ×0.01 PBS to obtain the transport properties of the device after the biofunctionalization. The channel current increased from 89 nA to 97 nA, as the solution gating voltage is sweeping from −0.5V to 0.5V. The increasing channel current along with the increasing gating voltage indicates that the synthesized InN film was n-type, which is also consistent with the studies using sputtered InN thin film (Saito and Igasaki 2001). The carrier mobility degradation of the FET channel after biofunctionalization was also clearly observed. From Figure 4(b), it can be seen that the FET transport curve becomes flatter after the biofunctionalization, and the on/off ratio of the device decreases from 1.08 to 1.01. The carrier mobility degradation could be explained by the disordered potential induced by the direct binding of biomolecules(Mao et al. 2010; Mao et al. 2013; Mao et al. 2011)

Detection of HIV gp41 antibody

The analytical performance of the device was evaluated using HIV-1 envelope gp41 antibody in PBS. The development of rapid and accurate diagnostic tools for POC diagnosis of HIV is of paramount importance for effective control of HIV infections (Davis et al. 2013; May 2017). The HIV-

1 envelope gp41 antibody is one the most commonly used HIV markers (Chin et al. 2011). To achieve specific detection of the gp41 antibody, 10 μg mL^{-1} of HIV gp41 antigen was used to bind with the APTMS linkers on the InN surface. The HIV-1 gp41 antibody was diluted with 0.01× PBS solution into different concentrations, and then tested using the developed InN RUM-FET biosensor.

Representative FET curves obtained from the HIV-1 calibration experiments are shown in Figure 6c. It can be observed that the source-to-drain current decreased with the increasing concentration of the HIV-1 gp41 antibody, which is the typical response of a FET biosensor. The n-type characteristics of the rolled-up InN FET biosensor contributes to the reduced channel conductance with increasing antibody concentration. As the gp41 antibody has a mildly acidic isoelectric point (pI) of ~6.9, it is negatively charged in the 0.01× PBS solution with the pH value around 7.4 (Gallerano et al. 2015). Therefore, the attachment of a negatively charged molecule to the InN channels is equivalent to a negative potential gating, which leads to a reduced electron density and thus electrical conductivity of the InN channel. The calibration results for detection of gp41 antibody in 0.01× PBS are shown in Figure 6(d), which was fitted in an S-curve based on the Hill equation. The LOD of the InN RUM-FET biosensor was determined to be 0.1 ng mL^{-1}, which is 20 times lower than that (2 ng mL^{-1}) of the standard ELISA kit. As indicated in the supplementary material, the device variations were mainly induced by the biofunctionalization of InN. By optimizing the biofunctionalization step, we envision that the LOD could be further improved. In our future work, we will also fully characterize the electronic characteristics of the InN microtube FET, which could provide useful guidance on optimizing the FET design for biosensing performance.

Compared to other existing rolled-up microtube biosensors, our device demonstrates the detection of real disease biomarkers with an acceptable LOD. In one of the most similar studies that reported a rolled-up InGaAs/GaAs FET biosensor, only the detection of organic solvents was demonstrated. The detection of clinically relevant biomarkers. such as cells (Bausch et al. 2017; Smith

et al. 2011), nucleic acid (Medina-Sánchez et al. 2016), and proteins (Yu et al. 2014), has also been demonstrated using other types of rolled-up microtube-based sensing mechanisms including optical ring resonator (ORR), electrochemical impedance spectroscopy (EIS), and self-propelled micromotor-based sensing. The ORR sensing, though highly sensitive, requires an expensive micro-photoluminescence microscope, making it less practical for use at the point of care (Smith et al. 2011). The rolled-up EIS biosensor shows ultrahigh sensitivity capable of detecting as low as 20 aM of avian influenza virus (Medina-Sánchez et al. 2016). The EIS and FET-based biosensing mechanism both share similar merits such as high sensitivity, microfluidic integration, ease of electrical measurement, and thereby hold great potential for use in clinical diagnosis.

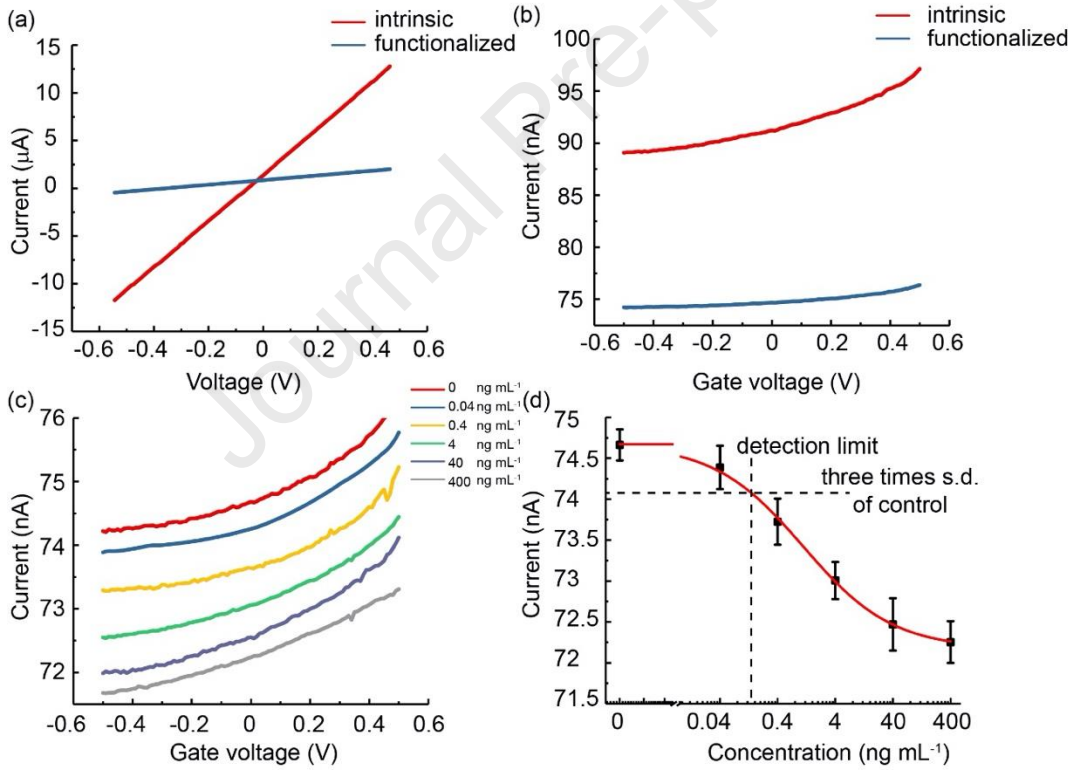

Figure 6. Electrical characterization of the rolled-up InN FET-type biosensor, and sensing of HIV gp41 antibody in PBS solution. (a) Source-to-drain current measurement results of the device before and after modification with HIV gp41 antigen (10 μg mL^{-1}). (b) Transport measurement results of the device before and after modification with HIV gp41 antigen (10 μg mL^{-1}). (c) Representative transport curves in response to HIV gp41 antibody at increased

concentrations. (d) Calibration results of rolled-up InN FET biosensor sensing results of HIV gp41 antibody, with data fitted into an S curve (N = 3).

Conclusions

In this paper, we presented the development of the microfluidic InN RUM-FET biosensor for label-free detection of HIV antibody. The rolled-up InN microtube was successfully fabricated with a high yield, benefiting from the robust and reliable microfabrication process. The sputtered InN film was characterized using the various characterization tools, demonstrating the successful synthesis of high-quality InN films. The developed InN RUM-FETs, with an external reference electrode, were integrated into microfluidic channels for convenient fluid handling and reduced sample/reagent consumptions. Using HIV envelope gp41 antibody as a model analyte, we achieved a LOD of 01 ng mL^{-1}, which is 20 times lower than that of standard ELISA. Compared to other existing rolled-up microtube biosensors, our device demonstrates the detection of real disease biomarkers with an acceptable LOD. In one of the most similar studies that reported a rolled-up InGaAs/GaAs FET biosensor, only the detection of organic solvents was demonstrated. The detection of clinically relevant biomarkers. such as cells (Bausch et al. 2017; Smith et al. 2011), nucleic acid (Medina-Sánchez et al. 2016), and proteins (Yu et al. 2014), has also been demonstrated using other types of rolled-up microtube-based sensing mechanisms including optical ring resonator (ORR), electrochemical impedance spectroscopy (EIS), and self-propelled micromotor-based sensing. The ORR sensing, though highly sensitive, requires an expensive micro-photoluminescence microscope, making it less practical for use at the point of care (Smith et al. 2011). The rolled-up EIS biosensor shows ultrahigh sensitivity capable of detecting as low as 20 aM of avian influenza virus (Medina-Sánchez et al. 2016). The EIS and FET-based biosensing mechanism both share similar merits such as high sensitivity,

microfluidic integration, ease of electrical measurement, and thereby hold great potential for use in clinical diagnosis.

CRediT authorship contribution statement

Pengfei Song: Conceptualization, Writing-Original Draft, Writing-Reviewing and Editing. **Hao Fu:** Validation. **Yongjie Wang:** Validation. **Cheng Chen:** Investigation. **Pengfei Ou:** Investigation. **Roksana Tonny Rashid:** Validation. **Sixuan Duan:** Writing-Reviewing and Editing. **Jun Song:** Conceptualization. **Zetian Mi:** Conceptualization. **Xinyu Liu:** Conceptualization, Writing-Reviewing and Editing.

Declaration of competing interest

The authors declare that they have no known competing financial interests or personal relationships that could have appeared to influence the work reported in this paper.

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