Printing Sensors on Biocompatible Substrates for Selective Detection of Glucose

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Abstract—This article presents development of organic electrochemical transistor (OECT) on a cellulose based biocompatible substrate for selective detection of glucose. The cost-effective manufacturing through drop on demand inkjet printing and blade casting techniques are adopted for the development of OECTs at ambient environment. The OECT is designed by printing carbon-based nanocomposite as the source, drain and gate electrodes, whereas for the channel layer, Poly (3, 4-ethylenedioxythiophene) doped with poly(styrene sulfonate) (PEDOT:PSS) is printed. Optical and physical characterizations were performed to investigate the printability, patterns' uniformity, repeatability, and adhesion of the printed structures onto the target substrate. The sensors are tested against specific enzyme i.e. glucose oxidase mixed in a PBS (phosphate buffer saline) solution. Transistor characteristics of the OECTs are determined to find the best coupling parameters for the device test. The optimal coupling parameters i.e. Vds at 0.8V and Vgs at 0.7V produced the transconductance i.e. 0.15 mS, in suitable ranges of the device test. Five different concentrations of glucose were tested in the electrolyte solution starting from the 1µM and at maximum 100mM, producing a prominent change in the current response i.e. 20% of its initial value. The distinctive current responses for each glucose concentration at an average response time of ∼15 sec show very promising results for rapid glucose detection. The low-cost fabrication of these printed sensors on biocompatible substrates show favorable results towards deploying these sensors on human bodies for real-time glucose monitoring.

Index Terms—Printed electronics, sensors, OECTs, flexible, biocompatible, organic materials.

I. INTRODUCTION

P RINTED electronics have attracted significant interest in recent years and are considered as key enablers for exciting developments especially in the area of flexible and wearable sensing applications. Fabrication of various functional layers of an electronic device has become much easier task with direct printing technologies [1]–[4]. The additive manufacturing through drop-on-demand inkjet printing has reduced processing steps that makes it an attractive approach for making low resolution devices [5]–[7]. The lower costs of printing systems and materials are the main driving forces for adoption of these techniques by the academic researchers as well as in the industry. [5], [8]. Besides the lower prices of printable materials, the location-specific deposition contributes further in lowering the processing cost due to less material wastage [9]. A wide variety of functional materials in their solution form can be processed and deposited on a broad range of flexible, foldable, and stretchable substrates [5], [7]. Many electronic devices have been realized so far as proof of concept and have been applied in real-time applications [10]–[16]. Amongst these, biosensors remain at the heart of these developments, especially for real-time human health monitoring [17]–[19].

The non-invasive detection of various bioanalytes in human body fluids has triggered the development of wearable biosensors and to establish the techniques ideal for these specific applications [18], [20], [21]. Among the various analytes, glucose level detection is one of the widely explored for monitoring the health condition of a diabetic patient [22]–[26]. Conventional testing methods in the medical laboratories are costly, less accessible, and above all involves the collection of blood samples for each test makes the process time consuming and painful for the patients, especially the elder ones. Therefore, interest in the non-invasive and portable diagnostic tools have gained much momentum in recent years, resulting
in the development of wide variety of sensors [22]–[24], [27]. Biosensors based on organic materials are promising due to their low cost, printability as well as biocompatibility make them distinguished from their inorganic counterparts [24], [28].

The three terminal devices i.e. OECTs offer significant advantages for their simple geometric configuration. These devices are easy to manufacture and are able to detect wide variety of analytes selectively by using their respective enzymes [29], [30]. The precise sensing mechanism of OECTs and their development on larger areas provide an easy approach towards improved enzymatic detection. The lower operational voltages i.e. <1 V make OECTs ideal for portable, handheld, and wearable sensing applications [31]. The well-known biocompatible organic conductor i.e. PEDOT-PSS is traditionally used for the active channel, where the doping and dedoping phenomenon occurring in the channel layer in the presence of enzymes-rich electrolyte solution is exploited to detect target analytes [32]. OECTs using PEDOT: PSS as an active channel have already been demonstrated and many have tackled the problem of glucose sensing or tried to analyze sweat [27], [29].

Fig. 1 shows a schematic diagram of co-planar structure of OECT. The charge carriers are triggered to move in the electrolyte solution corresponding to the applied gate voltage (Vgs). This charge variation as a result of Vgs causes the doping or dedoping of the channel material and thus changing the current (Ids) between source and drain [28], [30]. The channel material is either functionalized with corresponding sensing material or respective enzymes are mixed in the electrolyte solution for selective detection of a specific analyte. The quantized change in current (Ids) is correlated to the proportional change in the analyte concentration. PEDOT-PSS, an organic conductor is commonly used as channel material in the OECTs’ construction [37], [38].

The holes dominant (p-type) PEDOT-PSS is normally-on device having a certain amount of current flowing between the source and drain of the device. In normal conditions, the charge balancing and doping part of the channel materials i.e. PSS makes it more conductive in the absence of a gate voltage (Vgs) [30]. However, when a positive voltage is applied at the gate, an increase occurs in the concentration of cations (Y\(^{+}\)). The exact amount of de-doping depends on the potential drop between the electrolyte and the polymer channel, which, in turn, depends on charge transfer reactions that take place at the gate electrode. Commonly in electrochemical sensors, the O\(_2\)/H\(_2\)O\(_2\) couple are often replaced with a fast-redox couple, i.e. ferrocene [bis (n5-cyclopentadienyl) iron] (Fc)/ferrocenium ion couple, to overcome the under-consumption issues of oxygen. This facilitating redox couple mediates the rapid transport of electrons from the reduced enzyme to the electrode terminal. However, no mediators are considered in this research due to the reason for exploring the reduced materials’ utilization presenting an acceptable range of glucose detection.
As shown in the representative redox reaction of Fig 1 between the enzyme and corresponding analyte, the generated electrons (e\(^{-}\)) move towards the gate electrode, which results in an overall reduction of the current (Ids) flowing between source and drain through PEDOT-PSS channel [23], [27]. Simultaneously, the PEDOT-PSS is reduced, electrons are injected in the channel material resulting into recombination of electrons and holes. The cations present in electrolyte solution partly move in the channel material to ensure the electroneutrality. Since holes are the majority charge carriers in PEDOT-PSS, the channel conductivity is decreased upon increasing the gate voltage. The corresponding change in current (Ids) is directly proportional to the concentration of respective analyte added in the electrolyte solution.

III. DESIGN, MATERIALS AND PROCESSES

A. Design

A co-planar design is selected for fabrication of the OECT, where all the three electrodes and channel material are patterned on the same surface. Effective area of the sensor is kept large enough to contain sufficient volume of the electrolyte solution for testing. OECTs are designed according to the previously reported similar structure [35], however this work explores the different aspects by using biocompatible materials and substrates, a step further towards realizing wearable human body worn sensors. The larger dimensions of the device i.e. few millimeters (mm) wide make it interesting for fabricating these sensors through nonconventional manufacturing techniques such as inkjet printing. The demonstrated OECTs have been reported to perform successfully with dimensions as 10mm long PEDOT-PSS channel at a 1mm spacing from the gate electrode. Fig. 2 shows schematic diagram of the OECT, representing all the geometric parameters. A containment well to keep the electrolyte solution from flowing out during test defines the active sensing area, as it would be placed such that to prevent contact between ionic solution and the contacting pads. Dimensions of the containment well yielded an active area of 8 × 6 mm\(^2\) and a thickness of 2 mm, that could contain a total volume of 0.096 mL.

B. Substrate

Substrates play an important role in the design of experiments and processing conditions as well as the final application scenarios are based on their physical, thermal, and electrical properties etc. Besides the various important parameters in adopting a polymer-based substrate for flexible electronics, most important is their glass transition temperatures (Tg). Tg of a substrate determines the suitable materials to be processed and sintered upon at the end of manufacturing process. Therefore, a wide variety of polymer substrates have been developed offering different set of properties for various applications. This research employs biocompatible and biodegradable substrates i.e. Clarifoil obtained from Celanese (UK). The Clarifoil substrates are cellulose acetate films, suitable for the biosensing applications due to their biocompatibility and degradability. The 50\(\mu\)m thick substrate was selected for the printing experiments. Substrate were cleaned with deionized (DI) water followed by a dehydration step. A UV ozone treatment was performed for 3 minutes before printing.

C. Electrodes and Channel Materials

Carbon is a biocompatible material and has been used for developing electrodes for biosensors by many researches [39]–[41]. Carbon black was purchased from Sigma Aldrich (product no 699624) and mixed in ethylene glycol at ~40 wt.%. The synthesized solution has a viscosity of ~25 cP, which is in the suitable range for doctor blade coatings. PEDOT-PSS is an intrinsic p-type material and is commonly used for creating transparent electrodes in various thin film devices such as OLEDs, solar cells, and phototransistors etc. PEDOT-PSS is a promising material available in the solution form at specific rheological properties suitable for inkjet printing. PEDOT-PSS has been procured from Sigma Aldrich synthesized especially for inkjet printing and was used without any further modifications.

D. Electrolyte Solution

Phosphate Buffer Saline (PBS) is the commonly used ionic solution for testing biosensors as it replicates the human body fluid. PBS solution is prepared in DI water in such a way that it matches the salts concentrations similar to those of found in the human body fluids. A weight per volume solution was prepared by mixing 0.8g of PBS in 42g of DI water. For selective detection of glucose, respective enzymes such as glucose oxidase (Sigma-Aldrich (G7141) at 0.1mg/mL is added to the PBS solution.

E. Printing Experiments

The widely practiced printing methods through contact-based and contact-less have provided the opportunity to process a large variety of organic and inorganic materials. Compared to the contact-based printing where desired structures are designed on the printing tools, contact-less printing offers more advantages in terms of rapid design changes, versatility, customization, and re-structuring of the whole manufacturing cycle in a much cost-effective way [1]. Among the various techniques, drop on demand inkjet printing and blade casting are prominent and widely reported, due to their ease in processing, handling of a wide range of materials and above all the lower principle cost of the system. Drop-on-demand inkjet printing is driven by a piezo actuation mechanism where the functional materials in their solution
Fig. 3 (b) shows schematics of the mask used for printing sheet. Spacing between the source and drain is kept at 10 mm. The mask was prepared through laser etching of a thick plastic reproducing similar structures in large batches. The shadow as it is more versatile, processing is simple and capable of forming the gate electrodes. The blade coating technique is advantageous for printing platen under vacuum.

Two coating layers of carbon were executed with an interlayer delay of 15 minutes, allowing a wet-on-dry deposition. After soft baking on the hot platen, samples were cured in oven at 60 °C overnight. Fig. 3(c) shows the fabricated devices after developing the containment well filled with an electrolyte solution.

**IV. RESULTS AND DISCUSSIONS**

**A. Printing Results: Physical and Electrical Characteristics**

Quality of the printed structures influences physical, mechanical, and electrical properties of the sensing devices. OECTs are solution-based testing devices, therefore physical adherence of the printed structures to the substrate need particular consideration. Optical micrographs of the printed device show less dimensional variations as shown in Fig. 3(c). The minor variations of few micrometers occurring only in the electrodes are negligible as the size of the device is large and have negligible effect on sensor performance. Scanning electron microscopy (SEM) images of the printed thin films of Carbon and PEDOT-PSS are shown in Fig. 4. Thin films were analyzed at different scales to ensure the uniformity and materials distribution of the printed materials.

Figure 4 (a & b) shows carbon black SEM image at 10 and 1 μm scales, respectively. Similarly, Fig. 4 (c & d) shows SEM images of PEDOT-PSS printed film at 1μm and 100 nm scales. The SEM micrographs for both the deposited materials show uniformly printed layers with no cracks after curing. Adhesion-loss test of the printed films to the polymeric substrate is of prime importance, as upon interaction with the liquid medium during device test, could lead to delamination of the printed structures. Especially in the case of PEDOT-PSS, which comes in direct contact with the electrolyte solution. Therefore, an adhesion-loss test was conducted consecutively after initial curing step of PEDOT-PSS followed by final curing after carbon coating. A scotch tape test was applied to check the attachability of both the films. The presence of cellulose allows partly absorption of the printed ink in the substrate bulk, which also plays significant role in enhancing the adhesion. The multiple printing steps and wet-on-dry deposition also enhanced adhesion of both the surfaces. Electrical properties of both the layers were recorded using a digital multimeter.
Electrical resistance of the PEDOT-PSS channel can be tuned by varying the number of printing layers. Herein, the number of layers is kept higher i.e. 6 in order to keep the conductivity in the desired range within 500 ohms. Despite curing at lower curing temperatures of 60 °C (compared to typical 120°C), the higher number of printing cycles and prolonged overnight curing produces the desired conductivities. Resistance of the PEDOT-PSS layer recorded was ∼410 ohms, whereas for the carbon-based electrodes, sheet resistance of 45 Ω/sq. was recorded.

B. Electrochemical Testing

The electrochemical measurements were performed using a semiconductor parameter analyzer. To confine the electrolyte solution in the effective sensing area with PEDOT-PSS channel layer, a silicone based rectangular shaped containment well was created, as shown in the Fig. 3 (c). Geometric parameters such as thickness, width and length of the containment well were kept as 2 mm, 6mm and 8 mm, respectively. The length and width were designed in such a way to cover the whole effective channel area as well as completely overlapping the pads to avoid direct electrical contact between the electrodes through ionic solution. This well was sufficient enough to contain a considerable volume (100 μL) of the electrolyte solution. An initial amount of 50 μL enzymatic solution (PBS:GOx at mixing ratio of 1:7) was added into the well. The saturated glucose solution was prepared and added into the electrolyte in a way to test six different dilutions, in the range of 1 μM to 100 mM. The sampling frequency is important to determine, as it allows sufficient time to record the changing data occurring as a result of relatively slower biological reactions. Therefore, a measurement delay of 0.5 sec was applied between each sampling event and Ids-Vds curve was determined according to these settings as well. The sweeping parameters for Ids-Vds curves i.e. Vgs were kept varying from −0.2 to 0.8 V with a 0.1V step, whereas Vds was swept from −0.8 to 0.3 V with a 0.1 step interval. This combination resulted in 11 data points of Vds being measured at a given value of Vgs and a total of 11 × 10 = 111 values, for a single Ids-Vds curve. Sensors were probed and data was acquired starting from dry channel, adding electrolyte, and subsequently adding specific glucose concentrations in the electrolyte solution.

C. Transfer and Transconductance Measurements

The transistor characteristics of the printed sensors were analyzed by obtaining Ids-Vds and transconductance curves. The Ids-Vds curves was analyzed by sweeping the drain voltage (Vds) and gate voltage (Vgs) at a specific frequency, whereas keeping the source grounded. Fig. 5 shows Ids-Vds response at different gate voltages (ranging from −0.2 V to 0.8 V) of transistor with a clearly reaching the saturation regime. The maximum drain current (Ids) in the range of 0.1 mA was recorded in the saturation regime at the peak gate voltage. Upon positive gate voltage, the change in conductivity of the PEDOT-PSS is minimal because of the lesser ionic reactions occurring at the interface of channel layer.
These results are in-line with the theory of the operation mechanism of the OECT where an increase in the gate voltage leads to the reduction in current flow in the channel due to the dedoping phenomenon of PEDOT-PSS and increase in drain voltage leads to higher currents till reaching saturation.

To determine the best performance combination from the coupling of Vds and Vgs, transconductance (Gm) of the transistor was computed. The Gm was calculated by using the standard formula i.e.

$$Gm = \frac{\delta Ids}{\delta Vgs}$$  \hspace{1cm} (1)

where \(\delta Ids\) is the rate of change of drain current at corresponding change in the gate voltage. For this measurement, the voltage on the gate Vgs was swept with a step 10 times smaller (0.01V) giving more than a 100 values per given Vds. The measured current values shown in Fig. 6, were used to determine the transconductance curve.

The transconductance values accelerate from moderate to higher values at corresponding increasing gate voltages as shown in Fig. 7 (a). The transconductance values were deduced by matching with the variations recorded in the Ids values shown in the graph of Fig. 6. It is evidenced from the plot in Fig. 7 (b), that combination of highest voltage Vds at −0.8V and Vgs at 0.7V gives the highest transconductance i.e. \(\sim 0.15\) mS. The deduced value of Gm is in the acceptable range for such devices. The transconductance value can further be enhanced by tuning the electrical conductivity of the PEDOT-PSS channel layer through post-printing treatments. Combination of these two coupling parameters were used for the electrochemical testing of the sensor against the different concentrations of glucose in the electrolyte solution.

**D. Glucose Testing**

The electrochemical response of OECT device was analyzed by exposing the sensor in direct contact to the electrolyte solution. The elastomeric well to contain the specific volume of electrolyte was filled and the sensor was probed to record the current response. Prior to testing the glucose concentration, an offset current was recorded with and without the electrolyte solution. Fig. 8 shows the current response of the sensor at respective testing conditions. It is clear from the graph that drain current increases significantly by adding the electrolyte solution. The ions present in the electrolyte solution interacts with the channel material and modulates the current at higher threshold as compared to the stable current level at dry condition. The rising and stabilization curve of the current response after few seconds determines that a stabilization period is required to reach the saturation regime before adding glucose in the electrolyte. The glucose was added in electrolyte solution such that five different molar concentrations within the range of \(1 \mu\text{M}\) to \(100\text{ mM}\) were achieved. Fig. 9 (a) shows response of the sensor in terms of current (Ids) increase with each corresponding concentration of glucose added into the electrolyte solution. Fig. 9 (b) shows normalized response of the sensor at each corresponding measurement.

Both the graph shown in Fig. 9 (a & b) show a clear distinction between each consecutive increment of glucose concentrations in the electrolyte solution. The change in current response is \(\sim 4\%\) compared to the initial stabilized values at corresponding minimal concentration i.e. \(1\mu\text{M}\). Similarly, for higher glucose concentration i.e. \(100\text{ mM}\), the drain current increases significantly by \(\sim 20\%\). These results are congruent to the working principle of OECTs as higher concentrations
Fig. 9. Sensors response against different glucose concentrations and their normalized response.

provide a higher chance to bind to the enzymes and leading to higher electrons exchanges at the channel. The sensor reaches its stabilization current initially in a short interval $\sim$2 minutes, however the interval between successive measurements is minimized till the point they reach saturation. The increase and slight decrease in the current response at each measurement step is caused by the equilibrium between glucose and enzymes present in the electrolyte solution. Ideally the sensor should reach the saturation regime without losses in the current peaks, however this would require the active channel layer to be functionalized with the corresponding enzymes. In this configuration, enzymes are mixed in the electrolyte solution and the glucose can easily unbind. This might be affecting the exchange of electrons between the two molecules and resulting into increase in current across the channel. Nonetheless, these slight variations are acceptable and have insignificant effect on the overall performance of the sensor. The sensor’s average response time is also determined i.e. 14 sec, by considering the 60% change in the recorded current as shown in Fig. 10. The response time of the sensor is also in acceptable ranges compared to the commercially available glucose sensors, which give response typically in 5-10 seconds.

Repeatability of the same sensor for multiple tests is ideally required especially when targeting wearable sensing applications. Therefore, the same sensors were undergone for repeated tests at similar testing conditions and very slight variations in the current responses were recorded. These were also confirmed by counter checking the intrinsic resistance of the PEDOT-PSS layer under dry conditions. The electrolyte solution from the containment well was drained out and let the channel material dries under ambient environment. A slight increase of $\sim$15 ohms in the initial resistance (410 ohms) of the PEDOT-PSS was recorded, which could possibly be contributed from the insufficient dryness of the layer or residual ions left behind in the channel layer. Five different sensors with similar geometric parameters were tested to replicate the sensors’ responses. Fig. 11 shows graph of the standard deviation for the five tested sensors showing negligible variations. The variations were caused by the slight variation in the base resistances, which were mainly contributed from their manufacturing through printing. Minimal deviations in the pattern width as well layer thicknesses through solution based printing are obvious, and therefore these characteristics ultimately influence the final results. Nonetheless, these variations

Fig. 10. Response time of the sensor after adding 1mM glucose in electrolyte solution.

Fig. 11. Standard deviation in responses of 5 different tests performed on a single sensor.
are minimal and can be overcome by proper selection of the printing materials as well as controlled experimental conditions.

V. Conclusion

An all-printed OECTs based glucose sensor is fabricated on biocompatible and degradable substrates. The processing parameters for inkjet printing of PEDOT-PSS were optimized aiming to pattern the conducting channel of OECT at desired geometry. An overnight curing was performed by keeping the printed structures at 60 °C resulting into an electrical resistance of 410 Ω. Blade casting of carbon-based paste was performed for patterning source drain and gate electrodes. Reliability of the printed structures was determined by investigating the optical and physical characteristics. The Ids-Vds curves and transconductance characteristics were measured to deduce the best coupling parameters. The electrolyte solution prepared by mixing PBS and glucose oxidase was contained in a well, interfacing the PEDOT-PSS channel and gate electrode. The optimal coupling parameters i.e. Vds at 0.8V and Vgs at 0.7V produced the highest transconductance i.e. 0.15 mS. The different concentrations of glucose were added to the electrolyte solution starting from 1 μM and at maximum 100mM. A significant change in the current response i.e. 20% of its initial value at lower response time of ~15 sec show very promising results. The sensor structure would further be modified, and functionalization of the active area would lead towards wearing these sensors directly on human skin.

References


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