

RESEARCH ARTICLE

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Human sweat-based wearable glucose sensor on cotton fabric for real-time monitoring

Anoop Singh, Asha Sharma and Sandeep Arya*

Abstract

In this work, a human sweat-based wearable sensor for real-time glucose monitoring has been fabricated on a cotton substrate after treating it with a two-step polymerization of pyrrole. The pyrrole-treated fabric was coated with solution of copper sulphate pentahydrate to grow Cu layer. The cotton/pyrrole/Cu fabric was treated with the solutions of copper acetate and manganese acetate to form Cu–Mn transition-metal alloy via electrochemical deposition technique. Results indicate that the developed sensor is reliable with glucose detection limit of 125 μM and 378 μM . In addition, the sensor output ranged between 50 and 400 μM glucose with coefficient of correlation, $R^2 = 0.983$, indicating a linear range of output current. The sensor's response is not significantly affected by interferents. The developed sensor is also validated on human sweat with satisfactory results.

Keywords: Electrodeposition, Sensor, Wearable electrode

Introduction

With the rise of global diabetes, the biggest challenge in the health care system is to develop a novel glucose monitoring system that overcome the limitations of the traditional invasive nature of commercial blood glucose meters which requires repeated fingertip blood. According to a recent report by the World Health Organization (WHO), (Vogt et al. 2020) around 422 million people worldwide have diabetes and the majority of these people are living in low- and middle-income countries. Also, the Sustainable Development Goals (SDGs) aim to free all countries (poor, rich and middle income) from diseases and ensure healthy lives for all at all ages. Therefore it is the need of the hour to develop economical, efficient, robust, affordable, eco-friendly and noninvasive glucose detection method that could provide continuous and real-time glucose monitoring and has a broad market application prospect. A variety of designs, such as contact lenses, eyeglasses (Sempionatto et al. 2017), tattoos (Bandodkar et al. 2013) and wristbands (Gao et al. 2016),

which directly touch with the skin and body of a human are successfully developed by several researchers for non-invasive analysis as wearable sensors. Hence, for wearable sensors the necessities of the substrates are stretchable, flexible, light, biocompatible and comfortable to human skin. For non-invasive sensor fabrication different substrates which have been used are textile (Promphet et al. 2019), paper (Anastasova et al. 2017), and synthetic polymer (Martín et al. 2017). Specially, for monitoring the individual's physiological state great advances in wearable biosensors are reported (Sharma et al. 2021). Electrochemical sensors among the different types of wearable biosensors have received particular consideration for monitoring vital signals by detecting physiological components in sweat due to their noninvasive nature, which benefits from ease of measurement and reduced skin irritation (Bandodkar and Wang 2014). Many people suffer from diabetes which is a chronic disease originated by insufficiently produced insulin (Wang 2008). Glucose level of patient required to be continuously monitored because diabetes creates many critical complications. Normally, by analysing the glucose level of blood diabetes can be diagnosed. Also, during fasting the concentration of glucose in blood crosses 126 mg/dL and then

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hyperglycemia occurs (Ismail-Beigi 2012). However, it is exasperating to detect glucose level in blood as it involves bleeding with a needle. Moyer et al. (Moyer et al. 2012) successfully reported that glucose in blood correlates to the concentration of glucose in sweat. It is equivalent to 0.3 mg/dL glucose in sweat when 300 mg/dL glucose is present in the blood. Hence, it is probable that for diagnosing diabetes, the glucose level detection can be done through sweat sensor. Therefore, amount of the glucose level in sweat can give an important estimation to find out one's health conditions. For precisely measuring the pH and glucose level in sweat, the sensors should conformally touch the skin. Moreover, there should be good mechanical stability; otherwise, the sensor will deform because of the movement of the skin (Kim et al. 2011). In the body of a human, the circulation of metabolites and nutrients are the important indicators of biological processes, and their amount in bio-fluids can be used for prognosis, diagnosis, and clinical risk assessment and monitoring of therapeutic outcomes. Abnormality in the circulation of nutrients and metabolites (i.e. concentration) are directly related with health status such as cardiovascular disease and metabolic syndrome. Wearable sensor technology is crucial to developing personalized medicine by regularly examining the health condition of a person. It may be possible to have noninvasive monitoring enabled by obtaining a sweat sample from the human body, which is highly bioactive (Sonner et al. 2015). As an example, for diagnosing cystic fibrosis, the chloride concentration in human sweat acts as a gold standard and the concentration of glucose in sweat is mostly explored for management of diabetes (Lee et al. 2017). Human skin excretes sweat which is a biological fluid and can be utilized for non-invasive analysis. Sweat comprises metabolites (like glucose, lactate, uric acid and creatinine), electrolytes (like calcium, chlorides, potassium and sodium), proteins (like neuropeptide, tumor necrosis factors and interleukins) and small molecules (like cortisol, urea and amino acid) (Anastasova et al. 2017). For indication of the human health status, chemical concentration level in sweat can be treated as the biomarkers. Among the various biomarkers present in sweat, glucose is considered to be the crucial one as it can indicate diabetes status (Rosenberg and Kellner 1993). Nowadays, the human health is severely affected by diabetes and that's the reason it comes in the row of common chronic disease (Gupta et al. 2019). Diabetic patients have higher levels of glucose in their blood than normal and the symptoms it generated include kidney failure, nerve degeneration, blindness, neurological disorders, non-healing wounds, renal disease, cardiovascular disease and retinopathy (Witkowska Nery et al. 2016). Therefore, precise and fast detection of glucose

level is substantial in clinical diagnosis, biochemistry, individual blood glucose managements, food industry, etc. (Wang et al. 2018). Hence, urgent requirements to treat diabetes at very early stages attract much attention of the researchers in every year for fabricating sensors (glucose) with high stability, selectivity and sensitivity. The reason for fabricating enzymatic glucose sensor is great selectivity. For determination of glucose, enzymatic sensors such as modified glucose oxidase are extensively used (Lee et al. 2019). However, high cost and the poor chemical and thermal stability have put restriction on the commercialization of enzymatic sensors (Chaiyo et al. 2018). To overcome these disadvantages, it is worthwhile to build up non-enzymatic biosensor that causes glucose oxidation directly on electrode surface. In the past decade, some metals and their respective compounds were reported with enhanced properties as non-enzymatic biosensors (Muthumariappan et al. 2020). Moreover, some metals (like Pt) and alloys (like Pt–Pb alloy) used for fabricating non enzymatic sensors are known to be poisonous and expensive (Si et al. 2013). These problems can be solved by using transition-metal alloys and their oxides, such as CuO or Mn₂O₃ which exhibit very high sensitivity (Ci et al. 2014). Due to extraordinary properties, manganese-based mixed transition metal meets the requirement of the non-enzymatic sensor, because of the Mn atom having various valence states (Machini et al. 2013). Additionally, manganese material possesses various advantages such as high stability, facile synthesis, low toxicity and low price (Bohlooli et al. 2021). Some previously reported non-enzymatic sensor based on CoWO₄ micro-ring and CoWO₄ nanosphere possesses good electrocatalytic activity. However, these previously reported glucose sensor used rigid glassy carbon electrode for depositing CoWO₄ nanomaterials. Hence, for this reason it is very hard to fix these sensors on the human skin directly (Zhang et al. 2018; Singh et al. 2022; Park et al. 2003). In recent times, Kim et al. (Lee et al. 2016) have reported a chemical vapour deposition (CVD) and conventional photolithography methods for fabrication of conductive electrode-based stretchable electrochemical devices for monitoring diabetes. However, the developed sensor itself was not flexible, and for device stretchability, meandering interconnections were implemented.

The purpose of this study is to describe the creation of a stretchable electrochemical and skin-attachable sensor based on nanomaterials which is proficient of detecting glucose in human sweat. The fabrication of this electrode was done at room temperature with a simple immersion method and it avoids the usage of multistep and complicated processes, like photolithography and CVD. Additionally, for the modification of working electrode, the electrochemical deposition of copper was performed

because this deposition technique has various advantages such as control of the film thickness, low cost and simplicity (Lee et al. 2015). Hence, on the pyrrole deposited cotton fabric the Cu layer was grown successfully. Without use of an enzyme, the working electrode for glucose recognition was primed by the deposition of Cu-Mn alloy on top of cotton/pyrrole/Cu fabric via electrodeposition. No such work has been reported where pyrrole and Cu-Mn alloy-treated cotton fabric has been used as an electrochemical sensor for glucose sensing. The manufactured electrochemical sensor was directly applied to human skin without degradation and is extremely sensitive for detecting glucose in human sweat. Constant monitoring of the fluctuation in glucose levels caused by running and meal eating might be accomplished over an extended period of time using our skin-attached sweat sensor. Irrespective of the skin movements, the sensor was attached onto skin could be stably detecting values of glucose in sweat after exercising. These findings show that our sweat sensor has a great potential for use in personal physiological glucose monitoring when connected to the skin for a prolonged period of time. The ultimate goal of this work is to develop a non-invasive cotton-based electrochemical wearable sensor capable of detecting glucose levels in human perspiration. This technology can be worn directly on the skin or integrated into clothing to create a unique blood-free wearable sensor so as to monitor health.

Surface chemistry of cotton

Depending upon the diversity and augmentation conditions of the cotton fibres, its chemical composition varies (Hsieh et al. 1996). Every fibre of the cotton is made of layers which are concentric. The panoply of the fibre is itself detachable from the fibre and also comprises of pectin and wax materials. The primary wall is made up of cellulosic crystalline fibrils, which is the most secondary layer of the fibre. There are three different layers of the secondary wall of the fibre. The all three layers of secondary wall including tightly packed parallel fibrils have the spiral winding of 25–35° and also signify the cellulose inside the fibre. The lumen is the innermost part of the fibre of the cotton which is made up of remains of the contents of the cell. The lumen is packed with fluid consisting of protoplasm and cell nucleus before opening of the boll. Generally, cotton fibre mostly contains cellulose (88.0–96.5%) and some non-cellulosic components such as wax (0.4–1.2%), protein (1.0–1.9%), inorganics (0.7–1.6%), pectin (0.4–1.2%), and other (0.6–8.0%) substances. Except for colouring matter, protein and inorganic salts, which are located in the fibre lumen and rest of the non-cellulosics are positioned on the top layer of the fibres. During the growth and development of the

cell, the role of the non-cellulosic compounds is foremost and representing a barrier which is impermeable to the penetration of the liquor and they are also accountable for the yellow/brown colour of the native cotton. To facilitate effective succeeding wet processing of textile substrate (i.e. finishing and dyeing) (Topalovic et al. 2007), they are eliminated thoroughly by bleaching and scouring. The recent development in the modification methods of cotton surfaces has gained much interest of the researcher towards the sensing (electrochemical) of glucose.

Experimental sections

Chemicals

Sodium hydroxide (NaOH), pyrrole (C_4H_5N), Iron (III) chloride ($FeCl_3$), copper sulphate pentahydrate ($CuSO_4 \cdot 5H_2O$), copper acetate ($Cu(CH_3COO)_2$), manganese acetate ($C_4H_6MnO_4$) and acetic acid (CH_3COOH) all these chemicals are purchased from Thermo Fisher Scientific. Salts of Na_2HPO_4 (sodium phosphate dibasic dihydrate), NaH_2PO_4 (anhydrous sodium phosphate monobasic) and 5 mM of $[Fe(CN)_6]^{3-/4-}$ at different pH values were utilized for preparing 0.1 M phosphate buffer (PBS). For the preparation of PBS all the salts were also purchased from Thermo Fisher Scientific.

Methodology

Two steps are involved in the fabrication of a sensor that can detect glucose from human sweat.

Step 1, making cotton conductive

Firstly, cotton fabric was taken and washed with de-ionized (DI) water. Afterwards, it was immersed into the solution of 0.1 M NaOH for 3 h. Later, it was washed again with DI water followed by drying in an open environment. Then a two-step polymerization of pyrrole onto the cotton fabric is followed. In the first step, the dried scoured cotton fabric is immersed into solution containing 2 mL pyrrole and 100 mL DI at room temperature for 2 h. In the second step, the pyrrole-enriched cotton fabric is immersed into the solution of 1 g $FeCl_3$ in 30 mL DI water. $FeCl_3$ is used here as an oxidant such that it can initiate polymerization onto the cotton fabric at 20°C for 3 h. After that, wash the polymerized cotton with DI water. Dry the resulting cotton fabric in microwave oven at 50°C for 3 h. To enhance the conductivity of the pyrrole-coated cotton fabric, it requires a coating of highly conductive metal. 1 M solution of copper sulphate pentahydrate is formed for the electrochemical deposition on the pyrrole-coated cotton fabric. Maintain the pH of the copper sulphate solution in between 2 and 2.5 by adding few drops of sulphuric acid. In this way cotton fabric is

becoming highly conductive (shown in Additional file 1: Fig. S1).

Step 2, deposition of Cu-Mn on conductive cotton fabric

Firstly, two beakers were taken and 50 ml DI water was poured into each beaker. Afterwards, 2 g copper acetate was put in one beaker and in another beaker, 2 g manganese acetate was added and solutions were prepared. These two solutions were mixed under continuous stirring and 1 ml acetic acid was added into the resulting solution. The acetic acid plays a greater role to bind the metals on the conductive cotton fabric surface. The whole fabrication process is shown in Fig. 1.

Instrumentation

In order to conduct an electrochemical experiment, a three-cell configuration must be created on an electrochemical workstation. This involves the use of an Ag/AgCl reference electrode, a platinum counter electrode, and cotton/ppy/Cu/Mn modified electrodes as the working electrode. Cyclic voltammetry (CV) and differential pulse voltammetry (DPV) techniques have been utilized for studying the electrochemical response of unmodified and modified electrode. All electrochemical

studies were conducted using a 0.1 M PBS buffer solution comprising of 5 mM $[\text{Fe}(\text{CN})_6]^{3-/4-}$ over scan rate 0.05 V/s.

Characterization methodology

In order to gain a better understanding of the morphology, shape, size, and optical properties of the synthesized nanocomposite, the standard characterization techniques such as XRD (X-ray diffraction), environmental scanning electron microscopy (ESEM), and XPS (X-ray photoelectron) were utilized. A nanocomposite was synthesized and characterized using different techniques, including general characterization and some specialized instruments. Crystalline properties were studied using the Powder X-ray Diffractometer (Manufactured by Rigaku Corporation). In situ environmental scanning electron microscopy was utilized for morphological analysis.

Results and discussions

XRD characterization

The XRD patterns for the cotton and the pyrrole-coated cotton fabrics are shown in Fig. 2a. The X-ray diffraction pattern of cotton shows maxima at 2θ values of 14.8, 16.74, 22.87 and 34.43 assigned to (101), (101), (002) and (040) planes of cotton. The peak maxima of

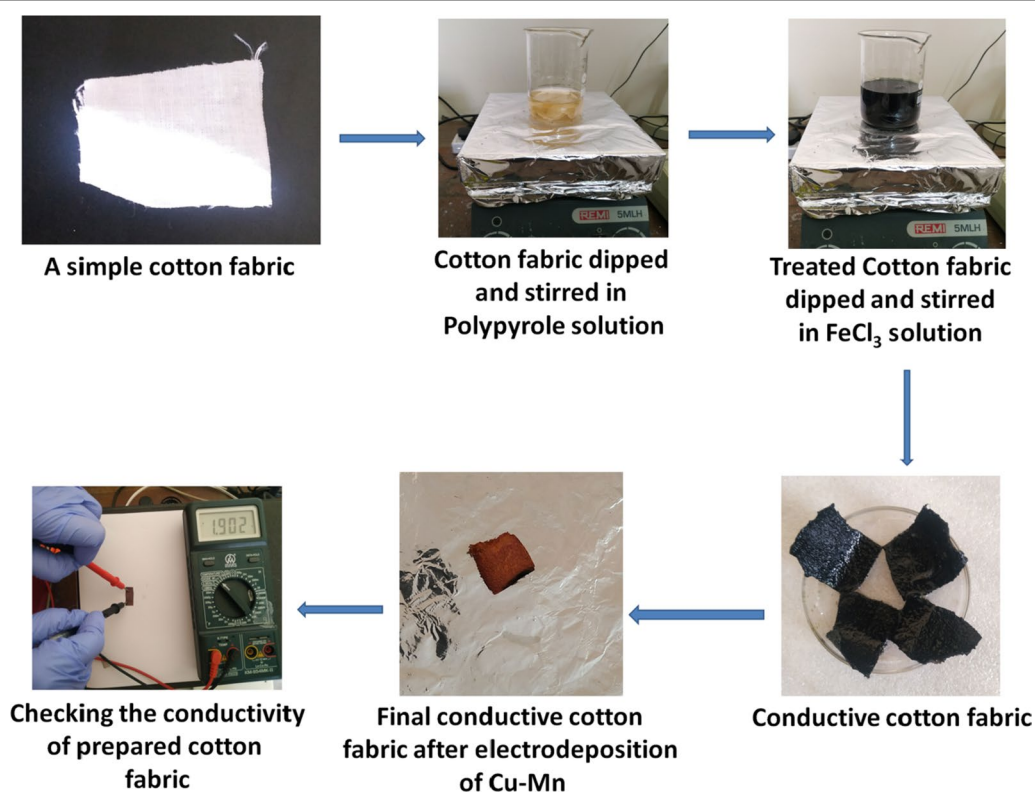


Fig. 1 Pictorial representation of steps carried out in the fabrication process

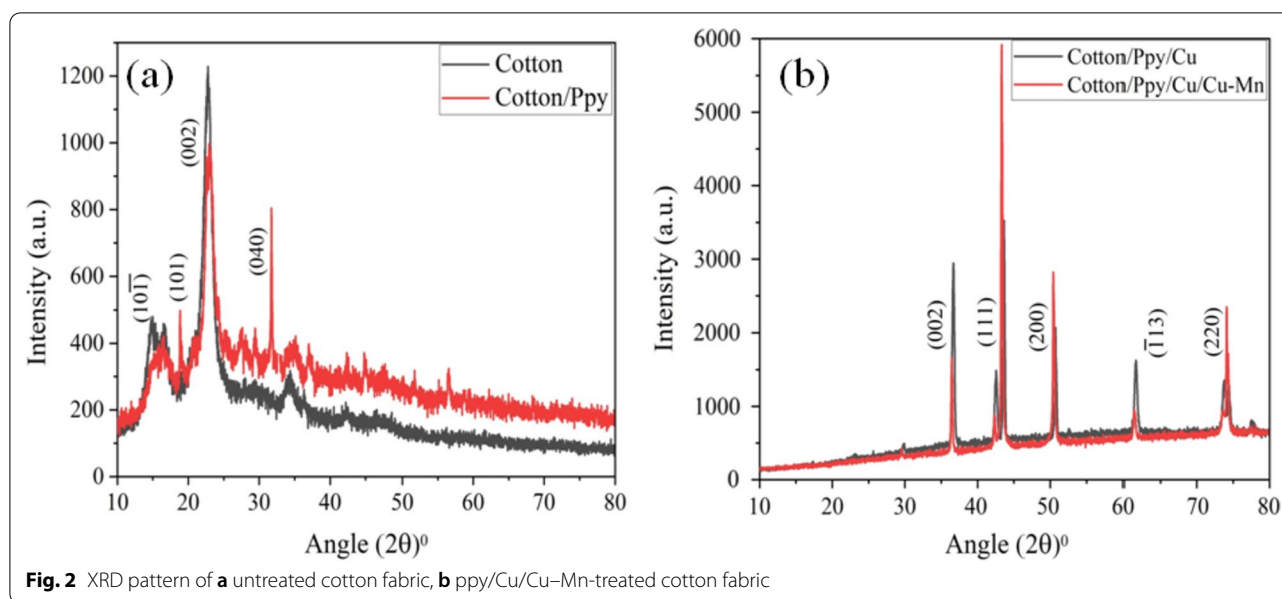


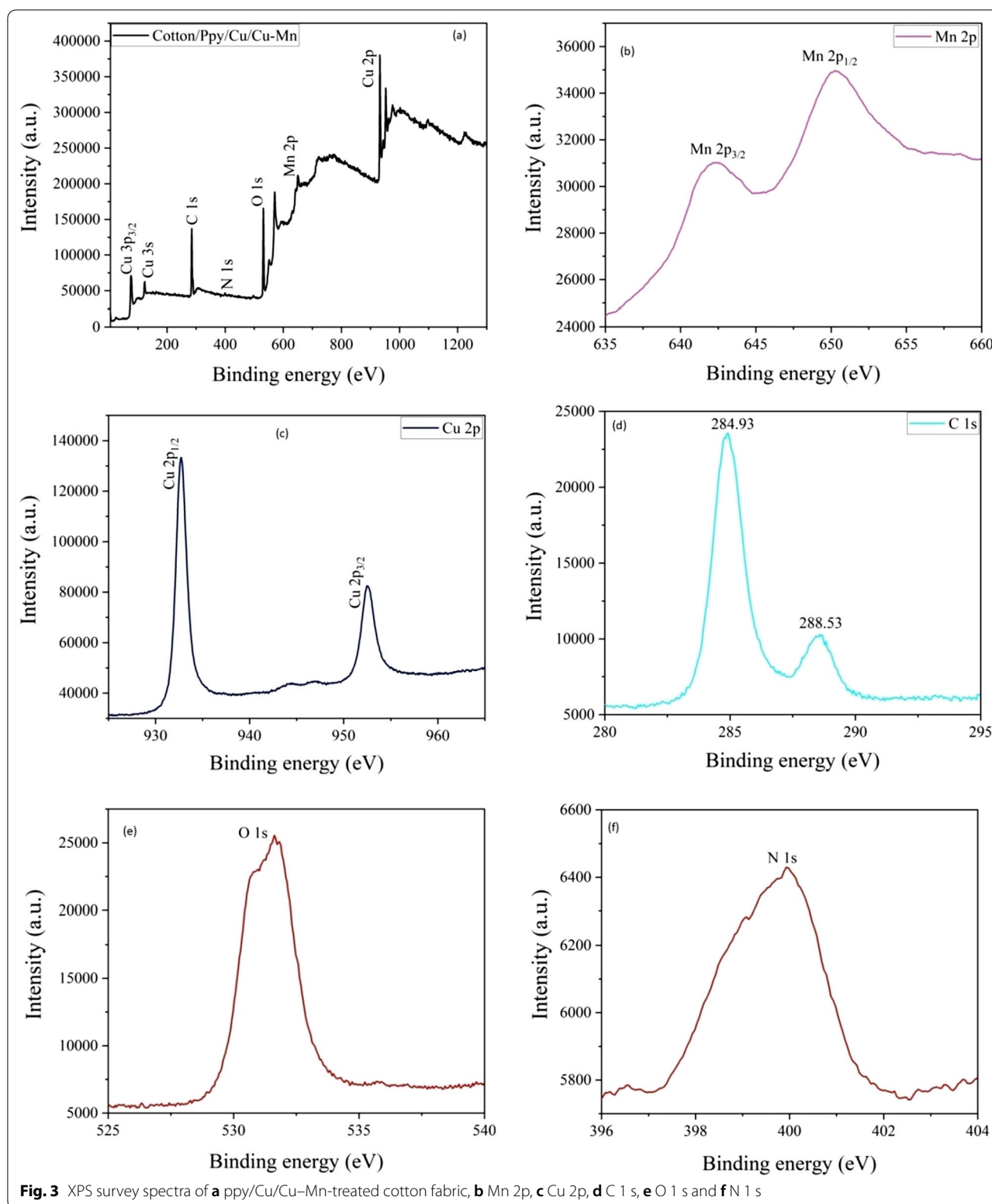
Fig. 2 XRD pattern of **a** untreated cotton fabric, **b** ppy/Cu/Cu-Mn-treated cotton fabric

pyrrole-coated cotton fabrics are found to be located at $2\theta = 16.19$ and 23.98 , and the values are slightly shifted towards higher side. From this, we can conclude that the pyrrole molecules were diffused into cotton fabric. In the pyrrole-coated cotton fabrics the peak at 23.98 has a hump like shape probably due to the presence of the polypyrrole. Further, additional peaks appear at 18.98 , 31.72 and 32.85 , corresponding to pure pyrrole crystals. This reveals the presence of pyrrole in the cotton fabric. The XRD examination was performed to validate the deposition of copper nanoparticles on the fabric (before electroplating). The XRD pattern for the 2θ range of 10° – 80° with a step of 0.02° is shown in Fig. 2b. All of the diffraction peaks are perfectly indexed to the copper structure, indicating their phase purity. The diffraction peaks at 43.66° , 50.72° , and 73.67° corresponded to the copper (1 1 1), (2 0 0), and (2 2 0) planes, respectively (Gan et al. 2008). The sharpness of the peaks shows that the copper particles are crystalline, whereas their widening implies the creation of nanoscale copper particles. Apart from the Cu_2O phase peak at 30.02° , 36.68° , and 61.72° , no impurity peak was identified (Zhao et al. 2013). The rise in the intensity of the XRD peak indicates the successful incorporation of manganese in the coated cotton fabric. The presence of manganese is further confirmed in the XPS analysis.

XPS characterization

The XPS technique can provide precise information about the composition of a material's initial atomic layers. XPS was used to validate the development of metallic copper and manganese on the surface of

pyrrole-coated cotton/ppy/Cu/Cu-Mn substrates. Carbon and oxygen are the only elements that can be analysed by XPS in pure cellulosic fibres, while hydrogen detection is not possible due to the technique's limitations. A low-resolution XPS scan is used to detect the oxidation states of the elements on the cotton fibre surface following pyrrole, copper, and manganese coating. The survey spectrum as demonstrated in Fig. 3a confirms the presence of carbon, oxygen, nitrogen, copper and manganese in the coated cotton fabric and the electronic state of the constituent elements are N-1 s, C-1 s, O-1 s, Cu-2p and Mn-2p. The four strong peaks at 285.0, 400.0, 531.0, 655 eV and 931 eV in full spectrum of coated cotton fibre were attributed to C-1 s, N-1 s, O-1 s, Mn-2p and Cu-2p, respectively. The chemical shifts of carbon (C 1 s) in cotton fibres can usually be easily classified into four categories (Mitchell et al. 2005): unoxidised carbon (C–C), carbon with one oxygen bond (C–O), carbon with two oxygen bonds and carbon with three oxygen bonds ($\text{O}=\text{C}=\text{O}$). After functionalization with pyrrole, copper and manganese the C-1 s XPS spectra of carbon with C–C bonding are shown at 284.8 eV and a minor peak at 288.5 eV (Fig. 3d) corresponded to C–N bond is also observed (Tang et al. 2007). Two distinctive peaks, with maximums of 642.1 and 653.6 eV, corresponding, respectively, to the Mn $2p_{3/2}$ and Mn $2p_{1/2}$ are found in the Mn 2p XPS spectra (Fig. 3b). XPS was also used to examine the chemical state of the Cu-coated cotton sample. The spectral graph in Fig. 3c shows the distinctive copper peaks. The peaks at 931.2 eV (Cu $2p_{3/2}$) and 951.1 eV (Cu $2p_{1/2}$) are associated with the Cu 2p



region of copper. XPS data validated the existence of elemental copper on the cotton's surface (Suryaprabha and Sethuraman 2017). The presence of an oxygen peak

was indicated by the value 531 eV in the XPS spectrum (Fig. 3e) (Kim and Lee 2014). The high binding energy

components at 400 eV can be assigned to nitrogen atoms of the pyrrole (Fig. 3f).

ESEM characterization

Imaging the surface of cotton fabric via environmental scanning electron microscopy (ESEM) showed distinct changes between the untreated and ppy/Cu-Mn-treated fabric (Fig. 4). As expected, the surface of the untreated samples showed a rather smooth and plain structure. In contrast, after treated with ppy/Cu-Mn, the surface of cotton fabric showed roughness, signifying the effective deposition of element on the cotton fabric surface.

Electrochemical studies

pH study

Primary research was conducted to determine the effectiveness of cotton/ppy/Cu/Cu-Mn modified electrodes

as an electrocatalyst in creating an effective surface for the analyte glucose and monitoring the conductive element's redox conversions. However, prior to conducting electrochemical studies, the pH effect on the surface of manufactured electrode was determined using PBS with a pH range of 7.0 to 9.0. The pH value of an electrolytic solution has a substantial effect on the electrochemical response of the sensor. Thus, the influence of pH on the electrochemical response of a cotton/ppy/Cu/Cu-Mn modified electrode in PBS was investigated using the CV technique over the potential range -0.75 V to $+1$ V (Fig. 5a). The peak current rose as the pH value increased and reached a maximum value at pH 8.0, at which point it declined. Thus, all electrochemical investigations were conducted in PBS with a pH of 8.0, demonstrating that the electron transfer mechanism occurs quite easily at this pH value.

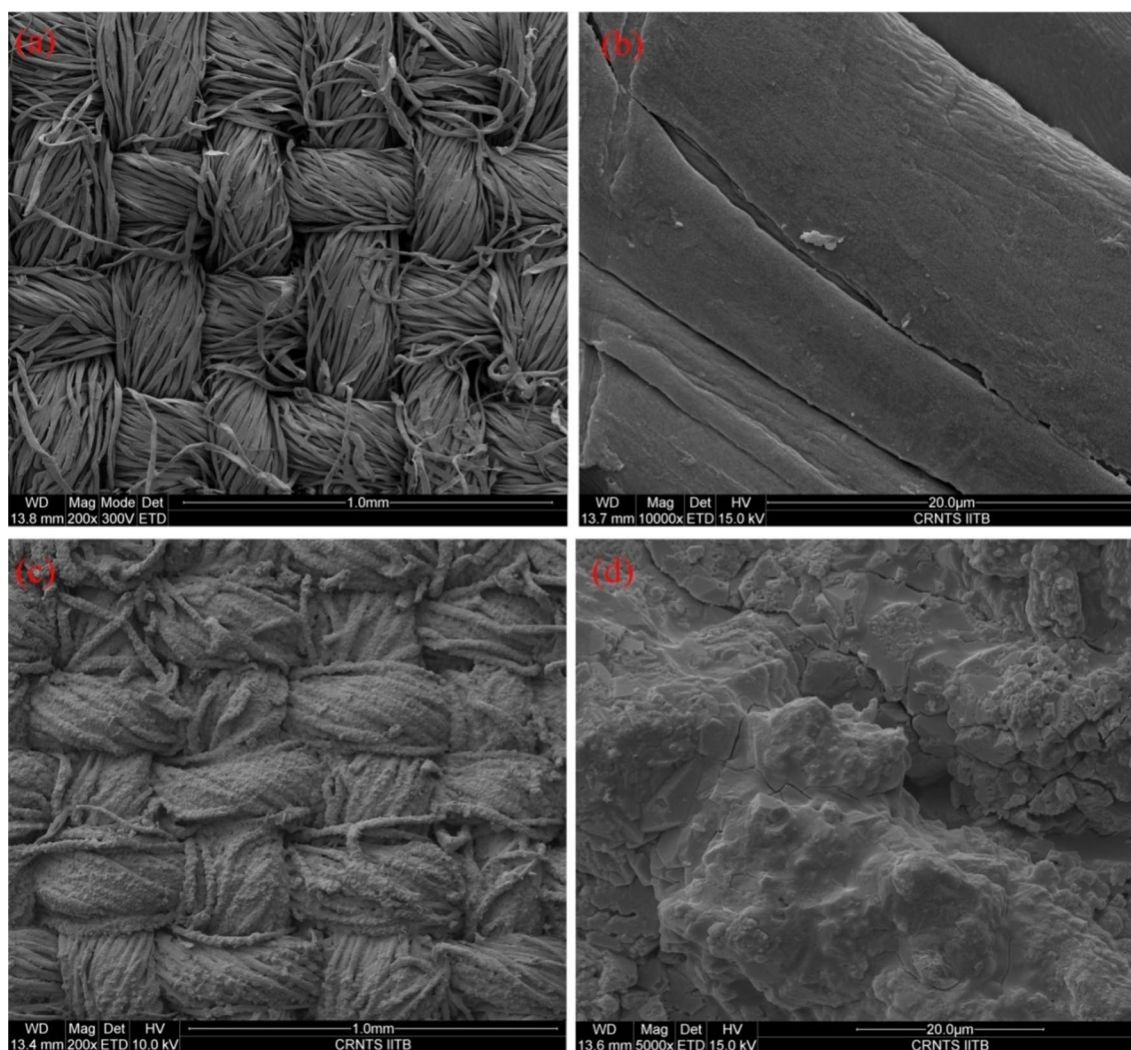
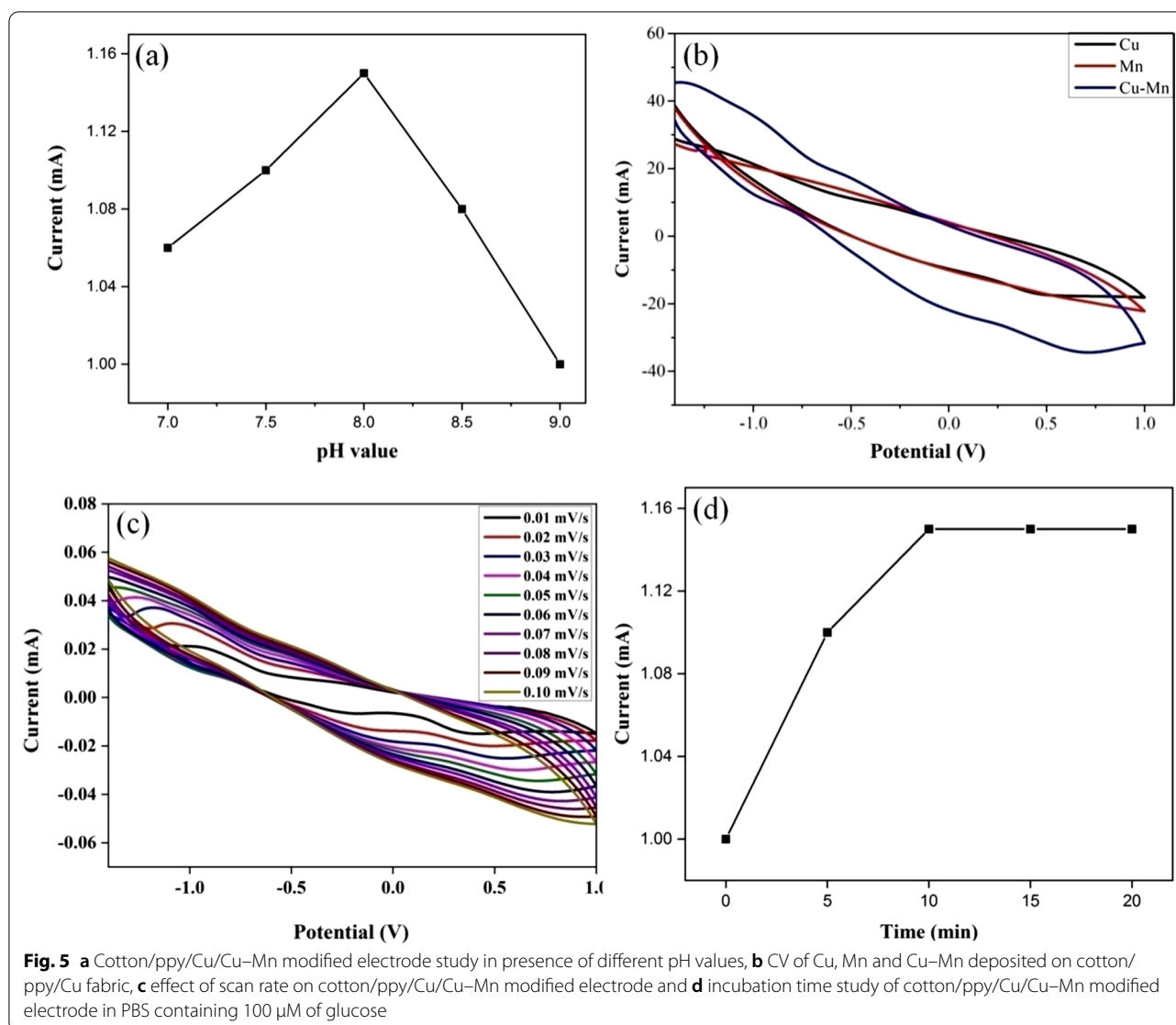


Fig. 4 ESEM images of (a, b) untreated and (c, d) treated cotton fabric



CV study

We investigated the electrochemical behaviour of the cotton/ppy/Cu, cotton/ppy/Mn, and cotton/ppy/Cu/Cu-Mn modified electrodes using CV methods in a PBS solution. This PBS solution was comprised of $[\text{Fe}(\text{CN})_6]^{3-/4-}$ in a concentration of 5 mM. All electrochemical experiments were conducted using a three-electrode cell configuration, with platinum wire behaving as a counter electrode and Ag/AgCl behaving as a reference electrode. CV was carried out by scanning the potential between -1.4 V and 1 V at a scan rate of 0.05 Vs^{-1} . Cotton/ppy/Cu and cotton/ppy/Mn electrodes had lower anodic and cathodic peak currents (I_{pa} and I_{pc}), while cotton/ppy/Cu/Cu-Mn modified electrodes had increased I_{pa} and I_{pc} values. As shown in Fig. 5b, cotton/ppy/Cu exhibited the smallest loop, which was enhanced for cotton/ppy/Mn, and an additional rise in loop area was observed for cotton/

ppy/Cu/Cu-Mn modified electrode. This increased current value indicates that electrons are being accretionary transferred between the cotton/ppy/Cu/Cu-Mn modified electrode and the electrolyte, resulting in greater electron diffusion between the cotton/ppy/Cu/Cu-Mn modified electrode and the redox species present in the electrolyte.

Scan rate study

To explore the electrochemical reaction's kinetics and behaviour on a cotton/ppy/Cu/Cu-Mn modified electrode, the scan rate impact on the peak potential and current at the electrode surface was investigated. The effect of scan rate from 0.01 to 0.1 Vs^{-1} on the electrode surface as shown in Fig. 5c was investigated using CV in PBS with a pH of 8.0 and comprising 5 mM concentration of $[\text{Fe}(\text{CN})_6]^{3-/4-}$. Each scan rate was associated

with distinct redox peaks. For the modified electrodes operating at 0.5 Vs^{-1} , the anodic and cathodic peak current ratio (I_{pa}/I_{pc}) was computed and was determined to be 0.989, which is roughly equivalent to one. This data indicated that the mechanism is based on quasi-reversible electron transfer with reversible kinetics for both electrodes. Additionally, it was established by the minor change in peak current values toward higher potential with increasing scan rate (Liang et al. 2013).

Incubation time study

The incubation time investigation (Fig. 5d) was conducted to determine the minimum time essential for the electrochemical reaction of 20 μL of glucose with a cotton/ppy/Cu/Cu-Mn modified electrode. The DPV approach was used to determine the electrochemical response of a cotton/ppy/Cu/Cu-Mn modified electrode towards glucose. Prior to performing glucose sensing at all concentrations, the incubation period was determined. The time interval between the addition of the analyte to the electrolyte and the development of a stable current following repeated application of the experimental procedure is called the incubation time. After electrochemical contact with glucose, the complex forms over the surface of the cotton/ppy/Cu/Cu-Mn modified electrode depend on the time and specificity of the modified electrode. The incubation time investigation was used to determine the response of the cotton/ppy/Cu/Cu-Mn modified electrode to a fixed glucose concentration (100 μM). The time was changed between 0 and 20 min and it was discovered that after 10 min, the magnitude of the current becomes constant; thus, 10 min was chosen for the electrochemical interaction of cotton/ppy/Cu/Cu-Mn modified electrode and glucose for response study.

Electrochemical sensing

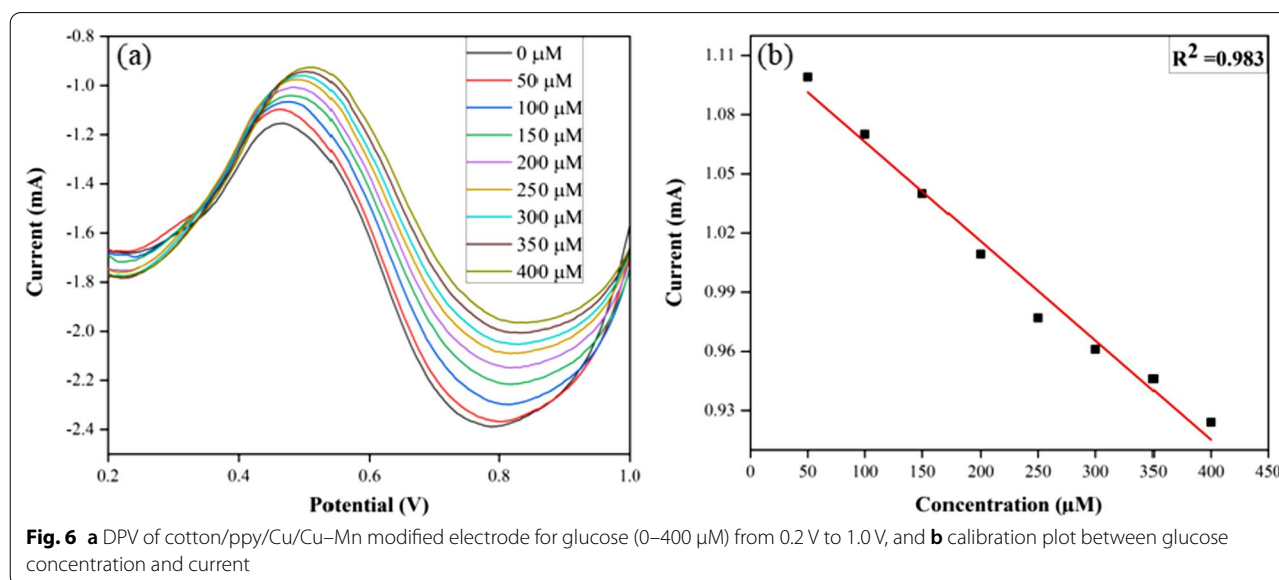
The electrochemical response of a cotton/ppy/Cu/Cu-Mn modified electrode was investigated utilizing the DPV technique for glucose measurement. DPV provides us with simply the faradic current and eliminates the capacitive current's influence on the current response. When electro-active species are oxidized and reduced, faradic currents are created. During the investigation, a ten-minute incubation period at 25°C is supplied prior to performing electrochemical measurements. Different glucose concentrations (0–400 μM) were investigated over a specific potential range (0.2 V–1.0 V) at 0.05 V/s scan rate. As illustrated in Fig. 6a, the current values decrease as the glucose concentration increases. This is because a concentration gradient arises at the electrolyte/electrode interface as a result of glucose reduction and oxidation occurring at the electrode surface. The thickness of the diffusion layer increases with time, resulting in a decrease in the concentration gradient, which in turn results in a reduction in the current.

Calibration plot of electrochemical sensing of glucose

The calibration curve between glucose at various concentrations and the peak current is shown in Fig. 6b. The coefficient of correlation R^2 is 0.982, indicating a high degree of linearity. The limit of quantification and detection abbreviated, respectively, as LOQ and LOD are determined from the following equations using the calibration curve slope, S derived from DPV and the standard deviation, σ of the minimum glucose concentration (Sharma et al. 2020).

$$\text{LOQ} = 10 \sigma / S \quad (1)$$

$$\text{LOD} = 3.3 \cdot \sigma / S \quad (2)$$

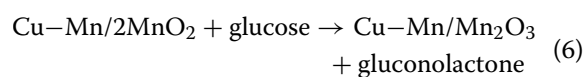
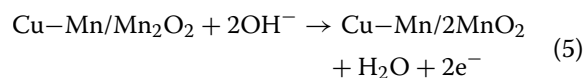
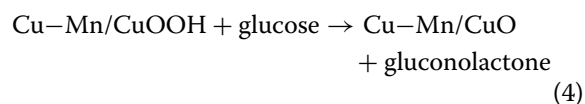
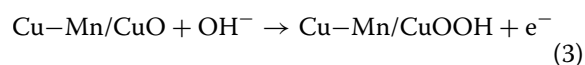


The detection and quantification limits of this synthetic electrode are determined to be 125 μM and 378 μM , respectively. The acquired result is also compared to previous studies, as indicated in Table 1.

Mechanism of sensing

Experimental evidence, in addition to spectroscopic characterization, supports the idea that the cotton/ppy/Cu/Cu-Mn modified electrode is sensitive, reliable, and resilient when used to detect glucose. While this may be true, the research found that Cu-Mn alloy materials had the ability to sense glucose. In general, such activity is done because of the presence of copper and manganese that do not interfere with, nor compete with, each other. A composite of these components yields a stable yet reactive electrochemical alloy electrode surface that can be used in the analysis of glucose. For the absence and presence of glucose, the electrochemical behaviour of Cu-Mn deposited on cotton/ppy/Cu was examined using differential pulse voltammetry, with testing occurring from 0 to 400 μM . When glucose was not present, the peak current of DPV is established in the PBS buffer solution as a result of the existence of cotton/ppy/Cu/Cu-Mn modified electrodes. The oxidation peak current reduces when glucose is added which suggests electrocatalytic oxidation of Cu^{2+} to Cu^{3+} oxidation. This is the same approach that Marioli and Kuwana (Liu et al. 2019) first proposed for copper electrodes. Glucose deprotonated and isomerized to produce an enediol, and subsequent adsorption on the electrode surface leads in an even more oxidizing Cu^{3+} state at the copper core. During DPV measurements, Cu^{3+} is primarily responsible for catalyzing glucose and producing gluconolactone, after which glucose acid is produced. Similarly Mn^{4+} is present in less quantity but it also converts the glucose into gluconolactone which further produced glucose acid. Manganese also enhances the stability of the fabricated electrode.

The diagram displayed in Additional file 1: Fig. S2 depicts the sensing mechanism stated above. The DPV profile of cotton/ppy modified by Cu-Mn alloy represents the typical diffusion-controlled electrochemical behaviour. This preliminary analysis concludes that glucose electrocatalytic oxidation occurs through the following four reactions (3–6) [52].

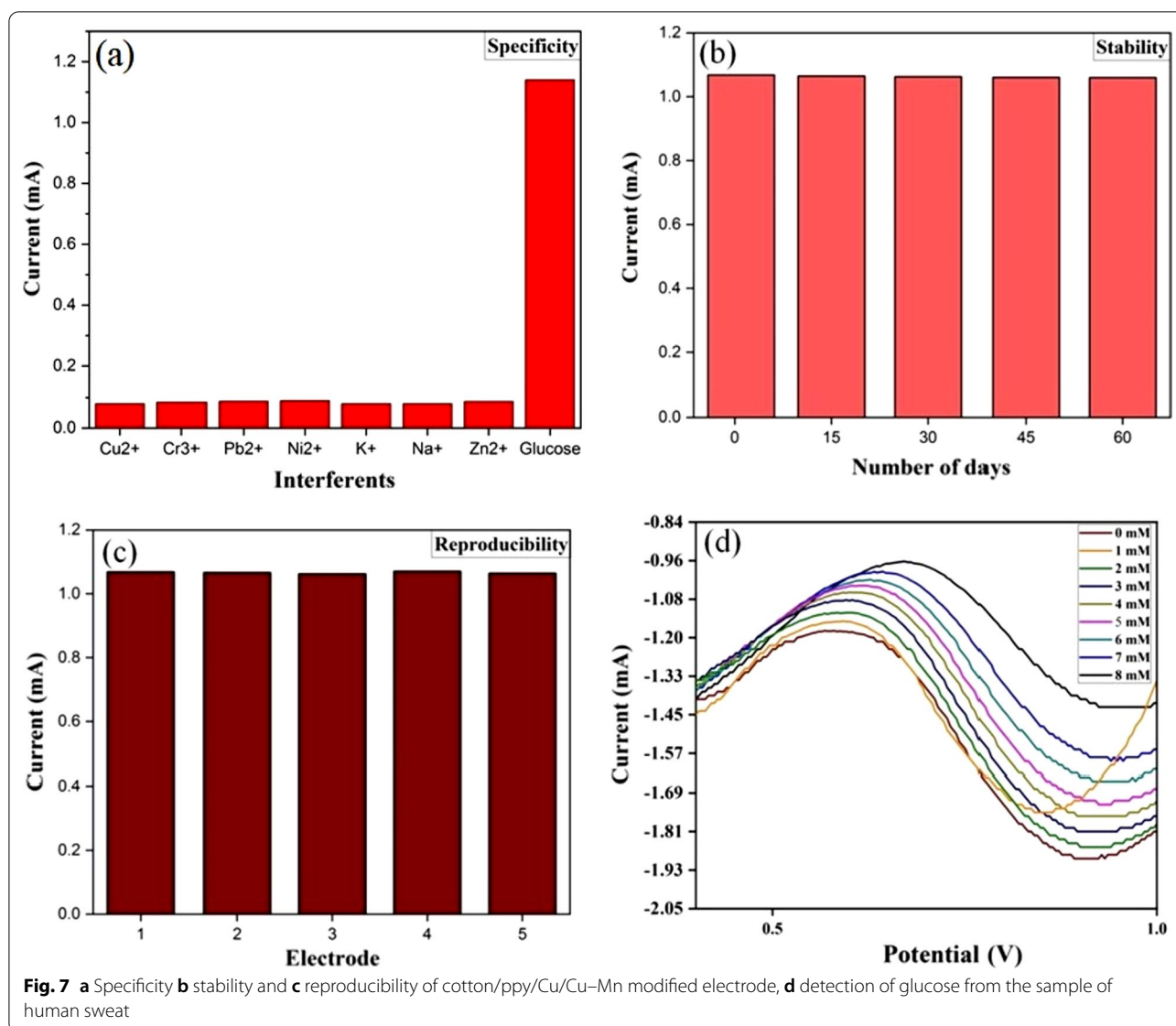


Specificity, stability, and reproducibility of the sensor

The developed sensing device exhibits a good electrochemical activity as well as stability. The created sensor's specificity was evaluated in the presence of additional interferences such as Cu^{2+} , Cr^{3+} , Pb^{2+} , Ni^{2+} , K^+ , Na^+ , and Zn^{2+} . Each interferent was tested at a concentration of 1 mM in the presence of 50 μM glucose. The interference investigation involved adding 20 μL of each of the above mentioned interferences to PBS. Prior to performing any electrochemical measurements, a ten-minute incubation period was provided. In the presence of interfering chemicals, no substantial variation in the peak current was found. The histogram of current with respect to interferences generated using the DPV is shown in Fig. 7a.

Table 1 Comparison of glucose sensor made from different electrode materials

Electrode materials	Linear range	Limit of detection (LOD)	References
Co_3O_4 nanoflower/graphene oxide hydrogels	0.25–10 mM	250 μM	Chung and Hur 2016)
Pt@carbon nano-onions	2–28 mM	90 μM	Mohapatra et al. 2018)
PB/Nafion/LOx	1–24 mM	1.31 μM	Gomes et al. 2020)
Au/Pt/Chit/LOx	Non-linear	1.0 μM	Currano et al. 2018)
Wearable cloth-based EC sensor	0.05 – 1 mM	–	Zheng et al. 2021)
ZnO NWs	0–25 mM	3.61 mM	Zhao et al. 2021)
N-doped porous carbon	5 μM –5 mM	–	Quintero-Jaime et al. 2021)
Silver nanowires	0.5–3442 μM	0.5 μM	Liu et al. 2021)
Ni–Co MOF/rGO/Polyurethane	10 μM –0.66 mM	–	Marioli and Kuwana 1992)
Cotton/ppy/Cu/Cu-Mn	0–400 μM	125 μM	This work



This demonstrated the electrode's selectivity for cotton/ppy/Cu/Cu-Mn. For a period of 60 days, the stability of the constructed electrode was determined by the variation in magnitude current values acquired from DPV (100 μ M glucose). When stored at room temperature, the modified electrode displayed excellent storage stability. The relative lifespan of the electrode was determined using differential pulse voltammetry to measure the response sensitivity of glucose on a periodic basis. If an electrode loses 10% of its initial signal, it was termed disabled. The results demonstrated that the modified electrode retained greater than 98% of its original signal after storing for eight weeks, indicating that the modified electrode remained stable even after 60 days. It was observed from Fig. 7b that the fabricated sensor demonstrated

excellent storage stability. Figure 7c illustrates the sensor's reproducibility. In the presence of 50 mM glucose and 500 mM interferences (as shown in Additional file 1: Fig. S3) response currents were almost identical to those induced by 50 mM glucose alone. This demonstrates the specificity of the modified electrode.

Real sample analysis

To demonstrate the feasibility of the manufactured electrode for glucose detection, the DPV method was used with a human sweat sample. The cotton/ppy/Cu/Cu-Mn modified electrode was evaluated for its real-world applicability via determining the peak current associated with various concentrations of human sweat (0–8 mM). For the real sample analysis, a fresh sample of human sweat

was employed. The current equivalent to 20 μ l of individual concentration (0–8 mM) has been determined using the DPV analysis. The DPV was determined by extracting 20 μ l of each concentration from human sweat. The DPV response of a cotton/ppy/Cu/Cu-Mn modified electrode with different concentrations of real sample is shown in Fig. 7d. For real sample, the peak current value corresponding to 1 mM is 1.15 mA. As illustrated in Fig. 7d, an increase in the concentration of the real sample results in a decrease in the value of current, which is accurately detected with the help of manufactured electrode. Additionally, the detection accuracy was unaffected through the presence of probable interferents in the human sweat sample. It implies that the constructed electrode is a good platform for sensing glucose from human sweat in the linear range 0–8 mM, even in the presence of additional components.

Conclusion

The main concern of humankind is the health of humans which drives the scientific community to build sensors capable of precisely monitoring and alerting to potential harm in real time. Clinically useful glucose sensors that are painless, do not require an insertion with needle, and run unobtrusively are required to advance the quality of life for diabetic people. In this research, a cotton-based sensor is fabricated for the non-invasive detection of glucose in human sweat. The wearable sensor fabrication is carried out in two steps (i) a two-step polymerization of pyrrole onto the cotton fabric in order to make it conductive (ii) deposition of Cu-Mn on conductive cotton fabric. The manufactured sensor employs electrochemical energy conversion as a transducing element for glucose monitoring in sweat, resulting in a self-powered, economical, and disposable glucose sensing podium for usage in a variety of applications. It was not necessary to use external power sources or complex external transducers with this sensor.

Supplementary Information

The online version contains supplementary material available at <https://doi.org/10.1186/s40543-022-00320-x>.

Additional file 1. Fig. S1. Detail analysis of the fabrication steps. **Fig. S2.** Demonstrate sensing mechanism of cotton/ppy/Cu/Cu-Mn modified electrode towards glucose. **Fig. S3.** Interference study with glucose.

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Authors' contributions

AS contributed to conceptualization, methodology, validation, and visualization; AS contributed to methodology, writing, reviewing and editing; SA contributed to conceptualization, supervision, methodology, writing, reviewing and editing. All authors read and approved the final manuscript.

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Availability of data and materials

The data will be available on a reasonable request.

Declarations

Competing interests

The authors declare that there is no conflict of interests or any competing interests.

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