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Woven Electrochemical Fabric-based Test Sensors (WEFTS): A new class of multiplexed electrochemical sensors

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We present textile weaving as a new technique for the manufacture of miniature electrochemical sensors with significant advantages over current fabrication techniques. Biocompatible silk yarn is used as the material of fabrication instead of plastics and ceramics used in commercials sensors. Silk yarns are coated with conducting ink and reagents before being handloom-woven as electrodes into patches of fabric, to create arrays of sensors, which are then laminated, cut and packaged into individual sensors. Unlike the conventionally used screen-printing, which results in the wastage of reagent, yarn coating uses only as much reagent and ink as required. Hydrophilic and hydrophobic yarns are used for patterning so that sample flow is restricted to a small area of the sensor. This simple fluidic control is achieved with readily available materials. We have fabricated and validated individual sensors for glucose and hemoglobin and a multiplexed sensor, which can detect both analytes. Chronamperometry and Differential Pulse Voltammetry (DPV) were used to detect glucose and hemoglobin respectively. Industrial quantities of these sensors can be fabricated at distributed locations in the developing world using existing skills and manufacturing facilities. We believe such sensors could find applications in the emerging area of wearables for chemical testing.

Introduction

Electrochemical sensors are used in a number of different analytical applications including tests for metabolites, blood gases, pollutants, pathogens and toxins[1-8]. The small form factor of the disposables and instrumentation along with the flexibility to test for a variety of different analytes using a range of techniques such as amperometry, voltammetry and impedance spectroscopy make them particularly amenable for point-of-care testing. Multiplexed detection of analytes using electrochemistry is especially useful to detect a range of analytes with a single sensor[9-14].

Currently, most commercial electrochemical sensors are multi-layer laminate assemblies with circuit elements fabricated using screen-printing on plastic or ceramic substrates and lamination bonded with fluidic elements like membranes or microfluidic plastic cartridges[15, 16]. Each electrode is printed sequentially through a separate screen followed by deposition of reagent where required. The manufacture of such sensors requires registration and alignment, the number of such steps increasing with the number of tests on the same sensor, necessitating the use of specialized fabrication equipment. Further, screen-printing results in significant wastage of electrode ink and reagent by nature of the technique[17]. Even with recent advances in paper microfluidics for electrochemical sensing[5, 18], fluidic elements must be integrated separately, which requires another layer to be bonded to the electrodes.

In this paper we demonstrate a textile weaving based approach for the manufacture of electrochemical sensors, which has several advantages over existing techniques. Previous efforts on textile electronics have demonstrated individual components like woven transistors[19] or assembled or screen printed sensors for humidity[20], temperature[21], pressure[22] or physiological parameters like ECG[23] onto patches of fabric. Recently, there have also been some reports on electrochemical sensing of metabolites like lactate[24] and glucose[25] using sensors made by screen printing on fabric. To the best of our knowledge, there have been no reports on using textile weaving itself as a scalable method for electrochemical sensor fabrication.

Yarns that have been pre-coated with conducting inks and reagents are woven into fabric at the required location to mass manufacture sensors in a single step. In contrast to screen-printed sensors, yarn coating can be done with exactly the required volume of reagents and inks resulting in lower costs and lower environmental damage. Also, silk being a natural material, we believe that the sensors produced using this

platform could be eco-friendly and reduce the generation of plastic waste from discarded or used sensors. Further, the weaving-in of electrodes into the patterned flow medium allows for the construction of simple one-layer sensors with fluid patterning created through the use of yarns with different surface properties. Multiplexed sensors can be easily created without the requirement of multiple registration and alignment steps by simply incorporating another electrode yarn into the fabric. There are few reported commercial multiplexed electrochemical sensors and we believe that our approach is particularly attractive to manufacture such sensors. Textile weaving is also widely practiced in many developing countries, has a low capital equipment requirement and is a simple yet highly scalable method to fabricate industrial quantities of sensors. A wide variety of yarn materials, weaving styles and looms provide the flexibility to tune sensor properties in ways that are not easily possible with a homogeneous medium like paper or nitrocellulose[26]. Advances in the automation of textile weaving can ensure accuracy and repeatability of sensors. Further, the emerging area of wearable[27-32] sensing devices will require the use of mechanically stable and reusable sensors, which can be embedded into clothing. Further, the well-known biocompatibility of silk can be used to interface these sensors with a human body in a wearable application. Textile materials provide the requisite combination of mechanical strength and functionality to be used in such applications.

As a demonstration of the capabilities of this new platform, fabric based glucose sensors were made by coating conducting ink and appropriate reagent on yarn to make different electrodes. We were able to detect physiologically relevant concentrations of glucose in whole blood with a maximum coefficient of variation (CV) of less than 5%. Apart from glucose sensors, we have also demonstrated the detection of hemoglobin using Differential Pulse Voltammetry (DPV). Finally, we have also shown multiplexed fabric based sensors for the simultaneous detection of glucose and hemoglobin using chronamperometry and DPV respectively.

Experimental

Material. Carbon ink and Silver/silver chloride ink were purchased from Creative Material Ink (USA). Carbon ink is selected to make the Working Electrode (WE) and Counter Electrode (CE) because it is inexpensive and inert towards unwanted species and atmospheric conditions. Ag/AgCl ink was used to make Reference Electrode (RE). Glucose Oxidase, (49180, 192 U mg^{-1}) from *Aspergillus niger*, potassium ferricyanide (III) (CAS-13746-66-2), sodium bicarbonate, D-(+)-Glucose and ethylene diamine tetracetic disodium salt dihydrate (EDTA) were purchased from Sigma Aldrich. Ammonium chloride (NH_4Cl) of 99 % purity was purchased from Rankem (India). Degummed (hydrophilic) and non-degummed (hydrophobic) silk yarns of 4 ply 25 twist per inch (TPI) were sourced from Silk Touch, Bangalore. An Em-Stat Potentiostat (Palm Sense, Netherlands) was used to perform all electrochemical experiments. A Nanophotometer (Implen, Germany) was used to take absorbance spectra of lysed blood sample.

Sensor preparation and use

Glucose sensors consist of three electrodes named Working electrode (WE), Reference electrode (RE) and counter electrode (CE). Carbon ink is coated using a custom-built yarn coating instrument and dried at 100^0C for 1hr to make counter electrode. For the working electrode, first carbon ink mixed with potassium ferricyanide (mediator, 1Molal) was coated and then dried in the same way as counter electrode. No degradation of silk fibers is expected or seen during the heating and coating process as they are protected by the conductive ink coating which can with stand higher temperature 170°C for more than an hour. Glucose Oxidase of concentration 2mg/100µL/ 50cm was then coated on mediator coated electrode. Ag/AgCl was coated on yarn to make reference electrode. The resistance of Working, Counter and Reference electrodes were 100, 50 and 0.5ohm/cm of electrode length respectively. Amperometric technique was used for detection of glucose and 5µL of blood was used to perform test. The design and dimension of the hemoglobin sensor is exactly same as glucose sensor except no reagent was coated on the working electrode and a carbon electrode was used in place of Ag/AgCl as the Reference Electrode. The resistance of electrode was 30Ω/cm. DPV technique was used for detection of hemoglobin and 5µL lysed blood sample was used to perform test. DPV voltammetry of lysed blood sample was performed with a scan rate of 50mV/s and pulse amplitude of 5mV. Peak current of different concentrations of hemoglobin were measured and correlated with concentration of hemoglobin. Sensors for the multiplexed detection of glucose and hemoglobin (Fig. 2e) consist of four electrodes in which three electrodes are made of carbon ink without any reagent on it and one working electrode (WE) of the same composition as used for glucose sensor. Electrodes were woven at 2mm distance and dimension of one sensor was 2cm X 1.2cm X 0.1cm. Sequential detection of glucose using amperometric techniques and hemoglobin by DPV technique was performed. RE and CE electrodes were kept common. Lysed blood sample of known glucose and hemoglobin concentration was used to perform detection of glucose as well as hemoglobin on same sensor.

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Sample preparation for glucose and hemoglobin detection

Calibration curves for glucose were constructed from samples made by spiking glucose in venous blood. Venous blood was collected in NAF (sodium fluoride) tubes, which helps in preserving glucose levels by inhibiting glycolysis[33]. Concentration of blood collected in NAF tube was measured using absorbance spectroscopy technique based instrument Bio system (BTS-350, UK), in Synergy diabetes clinic, Bangalore and was found 80 mg/dL. One stock glucose concentration of 600mg/dL was prepared by spiking glucose in blood. Spiked glucose was kept overnight at room temperature to equilibrate **α**

and **β** forms of glucose. Different concentrations of glucose were prepared by diluting stock concentration with the base blood (80mg/dL) concentration in proper proportion and the concentration was confirmed using instrument Bio system BTS-350. To prepare hemoglobin samples, blood was first collected in EDTA tubes. Plasma and leukocytes were removed by centrifuging the sample at 5000 rpm for 5 minutes at 4°C [34]. The remaining erythrocyte cells were washed three times with 0.9% NaCl. An equal volume of lysis buffer (8.04 g of NH4Cl, 0.84g of Sodium bi carbonate and 0.37 g of EDTA in 50ml of distilled water) was added to the cells and the entire mixture was centrifuged twice at 12000 rpm for 10 minutes to lyse the cells. Cell debris settles at the bottom of the centrifuge tube and the supernatant was collected as a stock sample. Absorbance spectra of lysed and diluted (0.1X) blood sample was measured in the range of 200 nm to 800 nm (Nanophotometer, Implen) and hemoglobin concentration was calculated using Beer Lambert's Law (Eqn. 1).

$$A(548) = \varepsilon * b * C \quad (1)$$

Where A is absorbance at a particular wavelength, ε is the extinction coefficient in cm^{-1} mM^{-1}, b is the path length in cm and C is concentration in of hemoglobin in mM. The value of ε used here was 12.39 cm^{-1} mM^{-1} and molecular weight of hemoglobin used was 64500g[35]. A stock concentration of hemoglobin (14 g/dL) was first made and diluted with distilled water to make different concentrations of haemoglobin.

Sample preparation for multiplexed sensor (glucose and hemoglobin detection). Sample for the multiplexed sensor was prepared in the same way as for hemoglobin. After lysis glucose was spiked in the hemoglobin sample and the stock was diluted with DI water, which resulted in dilution of both, hemoglobin as well as glucose. The same sample was then used for sequential detection of glucose and hemoglobin using amperometric and DPV technique respectively. Lysed blood sample of known glucose and hemoglobin concentration was used to perform detection of glucose as well as hemoglobin on same sensor.

Clinical testing of glucose sensors. We sourced 30 discarded clinical samples from Synergy Diabetes Centre (Bangalore, India). These samples were tested in triplicate on both our fabric sensors and Bio system BTS-350 using absorbance spectroscopy.

Results and Discussion

Design and fabrication of electrochemical sensors for glucose and hemoglobin. Most commercial sensors for blood testing are multi-layer assemblies consisting of 2-3 electrodes (working, reference and counter) screen printed on a ceramic surface and pressed into contact with a piece of membrane that wicks blood directly from the finger. To accomplish the same functionality, fabric sensors were made by the two simple steps of coating and weaving. Electrode yarns were prepared by coating of conductive inks (Fig. 1a) and reagents followed by subsequent drying. These electrodes were then woven into the fabric in the weft direction (Fig.1b&c) to make patches of sensors (Supplementary Video 1). Large numbers of sensors (Fig. 1d) can be woven as patches and are then pasted on medical grade adhesive backing for mechanical strength and then laminated to protect the sensor area. Individual sensors are then cut and tested (Fig. 2a&b) for the detection of the metabolite of interest using a potentiostat. For coating the yarns, we used a custom-made yarn coating instrument (Fig. 1a), which uses only as much reagents (conducting inks, enzymes and other chemicals) as required, reducing wastage and cost when compared with screen-printing
Arrays of sensors were covered with a laser-cut laminating film that provided a window for the application of blood and a separate window for making contact with a reader (Fig. 2a). Individual glucose sensors of size 2cm X 0.8cm X 0.1cm were cut out (Fig. 2b) before use.

Confining the flow path for blood in individual sensors was achieved by using a combination of degummed (hydrophilic) and non-degummed silk yarns (hydrophobic) in the warp direction. We were able to use readily available materials and simple boiling to make silk yarns with these dissimilar properties. In contrast to other natural and synthetic fibres, silk processing is simple and less chemically intensive for these patterning applications. The only processes involved are degumming to make silk hydrophilic which can be done either by chemical processing or even just by boiling. Blood sample flows along the hydrophilic yarn channel while not wicking into the hydrophobic background (Fig. 2c). This simple introduction of fluid patterning is in contrast to the complex multi-layer assemblies that are required in commercial sensors. Connecting points to interface the sensor with the reader are made by avoiding laying warp yarns at particular positions (Fig. 2c). A custom-made connector was used to interface the sensor to a potentiostat (Fig. 2d). Adjacent electrodes were separated by hydrophobic weft yarn woven with a thickness of 1mm. Both glucose and hemoglobin sensors had the same design with the hemoglobin sensor not having any reagent coated on the Working Electrode (WE) and with a carbon electrode in place of the Ag/AgCl electrode for the Reference Electrode (RE). Sensors for the multiplexed detection of glucose and hemoglobin (Fig. 2e) consist of only one more electrode than the glucose sensor. This fourth electrode is also made by coating carbon ink and serves as the Working Electrode for Hemoglobin with the RE and CE electrodes kept common for both hemoglobin and glucose. In contrast to previous work[18], only one electrode was added for this multiplexed detection, reducing sensor size and complexity.

Detection of analytes on fabric sensors. Electrochemical detection of glucose in blood was performed using chronamperometry. A sample volume of 5 µL was applied on the hydrophilic portion of the sensor. A constant potential of 0.5 V is applied across the working and reference electrodes and output current is measured as a function of time. An equilibrium time of 8s, after the application of the

voltage, was given for uniform distribution of sample (Supplementary Information S1) across the entire length of the hydrophilic channel (Fig. 2c). During this time, the sample become stagnant, eliminating convection based transport and the reaction at the electrode becomes purely diffusion controlled. The well-known combination chemistry of Glucose oxidase enzyme and potassium ferricyanide as mediator was used on the working electrode[14]. In the presence of enzyme, glucose molecules are oxidized to produce gluconic acid while potassium ferricyanide is reduced to ferrocyanide producing a current, which can be measured. This output current is directly proportional to the number of glucose molecules getting oxidized and decays with time according to the Cottrell equation [36]. Output current as a result of oxidation of glucose was measured with respect to time and the reading of current at 8s (which is equivalent to t = 0 s in the Fig 3a) after application of voltage was taken and correlated with the glucose concentration. The hydrophilic channel was designed in such a way that 5μL of blood sample was sufficient to wet the channel completely in order to avoid any variation due to changes in input sample volume. For a particular concentration of glucose, as the reaction proceeds, the local concentration of glucose molecules available for the reaction at electrode surface decreases with time and hence output current decreases (Fig. 3a) as per the Cottrell equation[7,37]. Increasing concentrations of glucose in samples show increasing value of current at the same time point (Fig. 3a). A calibration curve was constructed using this data (Fig. 3b) and served as the basis to report unknown concentrations that were later tested.

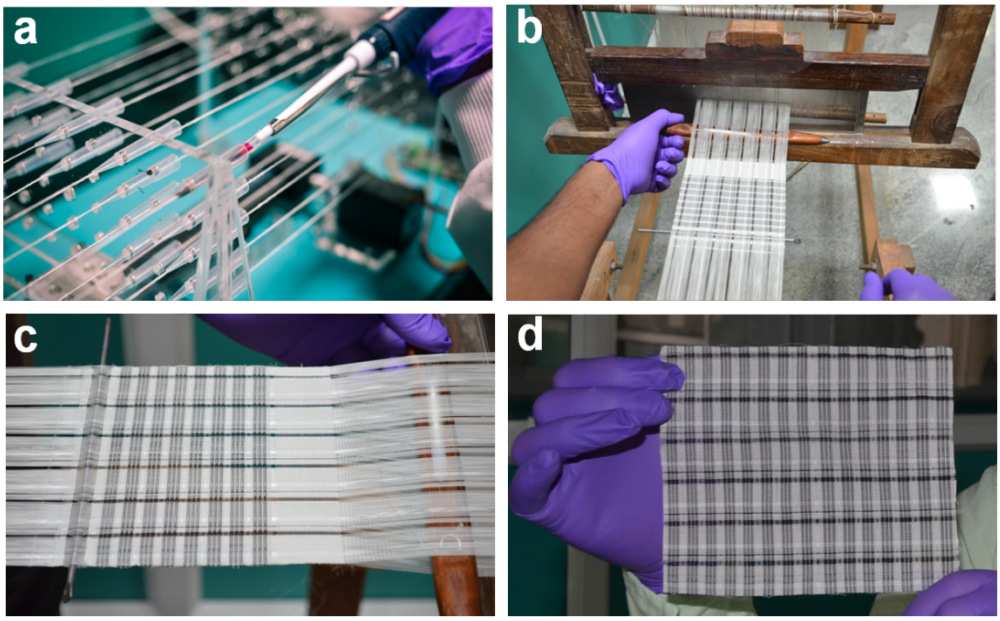

Fig. 1 Manufacturing of fabric-based electrochemical sensors. (a) Custom-made yarn coating instrument for coating conductive ink and enzymes and other reagents to make electrodes. (b) A view of the handloom used to weave the sensors. Weaving of electrodes in to the weft of the fabric was accomplished using a shuttle (wooden tool in the left hand of the weaver). **(c)** A closer view of woven patches on the loom which will be cut in to the individual strips after pasting on medical grade backing. (d) A woven array of 90 (15 X 6) sensors which was cut from the loom.

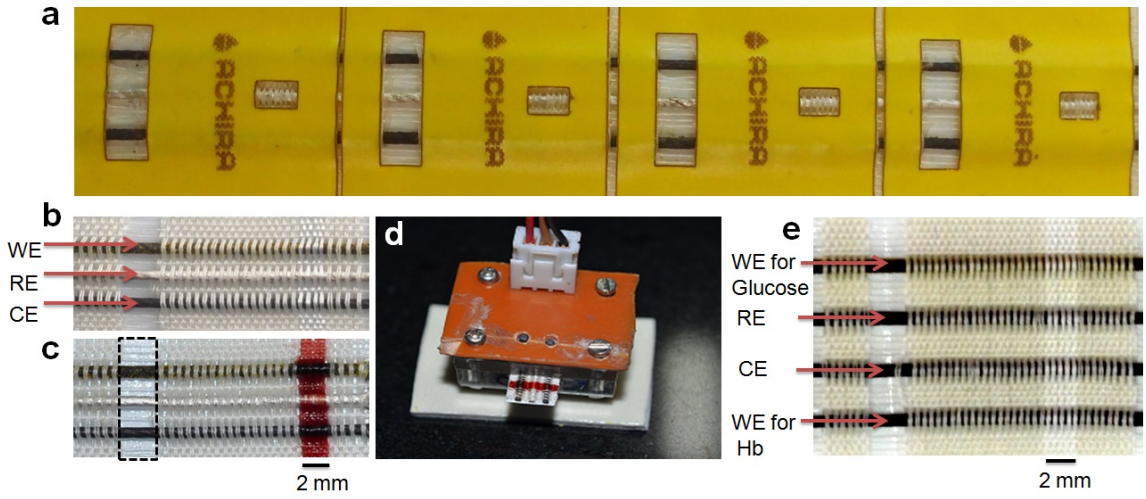

Fig. 2 Sensor design and interfacing. (a) Single packaged patch of woven glucose strip containing four individual strips and gap indicating the mark to cut individual strips, **(b)** Design of single glucose strip showing working (WE), reference (RE) and counter (CE) electrodes, **(c)** Glucose sensor showing connecting points for potentiostat using a dotted line where there are no yarns woven and also showing the path for the whole blood sample (red) along the hydrophilic channel made by combination of hydrophilic and hydrophobic silk yarn. **(d)** Fabric sensor interfaced with an adapter to read and perform the amperometric test for glucose sensing, (e) Multiplexed fabric sensor for the detection of glucose and hemoglobin. The same CE and RE are used for both tests.

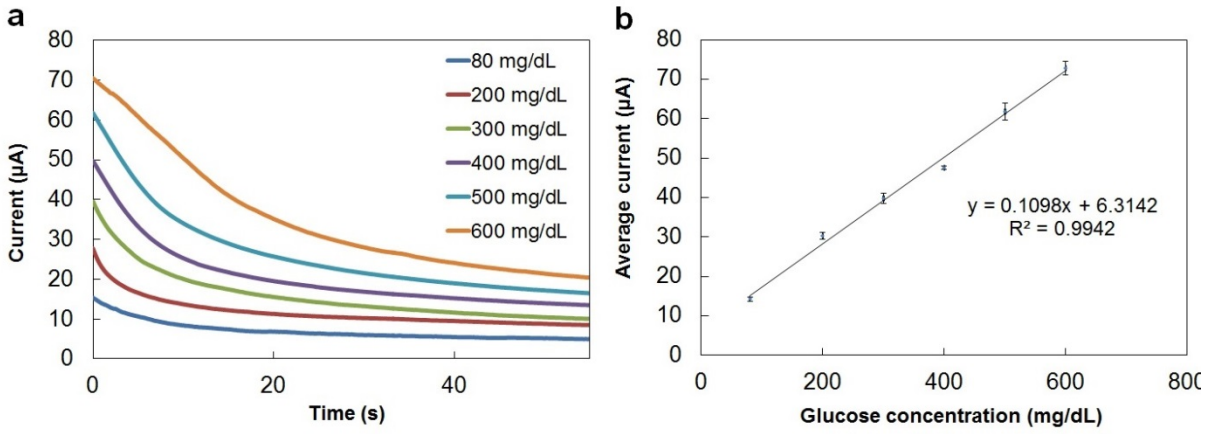

Fig. 3 Amperometric signal curves and calibration curve for glucose. (a) Amperometric profile of different concentrations of glucose spiked in the blood at a potential of 500 mV on the fabric sensor. For a particular concentration of glucose, signal decays with time as the glucose in the sample consumed. Increasing concentrations of glucose show increase in amperometric current. **(b)** Peak current values are used to construct a calibration curve. Three replicates at each concentration were used to generate caliabration curve. The curve shows good linearity for glucose detection with CV < 5% for individual values.

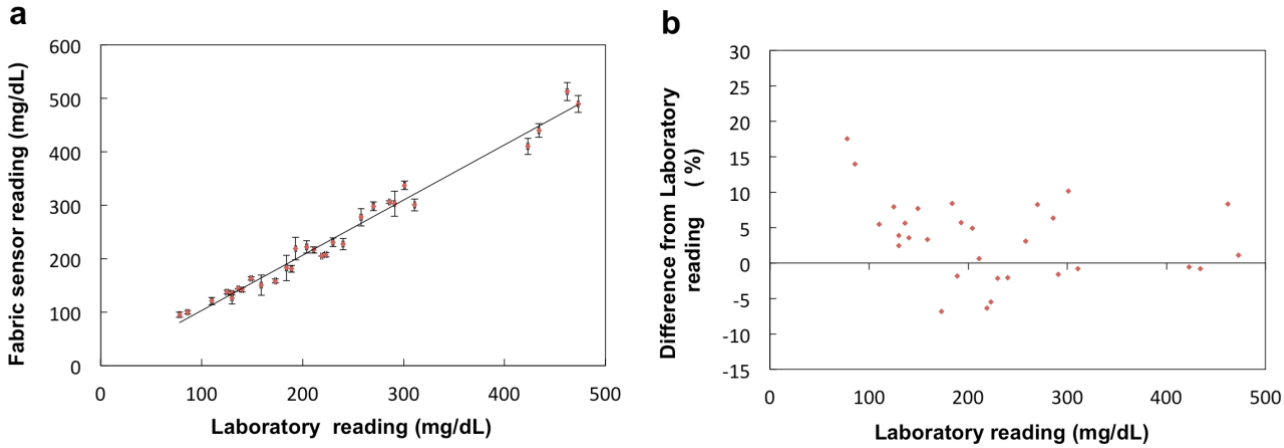

Fig. 4 Clinical evaluation of glucose sensors. (a) Accuracy of the fabric based glucose sensor was measured by plotting concentrations obtained using fabric sensors against the laboratory reading. Each concentration was replicated 3 times. The line y = x is shown as a guide to the eye. The best fit line was y = 1.03X + 0.04 and had an $R^2 = 0.98$. Standard deviation is shown for fabric sensors. A strong correlation was found between the laboratory reading and the fabric based glucose sensor (b) Plot of % difference between the fabric sensor and laboratory reading in the Bland- Altman format show that most of the fabric reading are within ± 15% of the gold standard

The glucose sensors showed very good linearity (Fig. 3b) in the range of 80 mg/dL to 600mg/dL which covers the clinically relevant range for glucose concentration in humans. The Coefficient of variation (CV) with repeat testing of samples on the same lot of sensors was less than 5%, which we believe can be further improved by introducing automation in the coating and weaving process. This low CV is also reflective of the uniform coating of ink and enzyme that is achieved on yarns through our coating process.

Clinical performance evaluation of fabric based glucose sensor

We have also tested the performance of fabric sensors using 30 live clinical samples to prove the clinical relevance of these fabric-based glucose sensors. First, an accuracy plot showing the reading from the fabric sensor was plotted against the laboratory reading (Fig. 4a). Very good correlation was found between fabric sensor and laboratory reading with $R^2 = 0.98$. The lowest concentration that was available from the clinical study was 78 mg/dL and the highest obtained was 480mg/dL.

The criteria of the U.S Food and Drug Administrator (FDA) for a glucose sensor to be clinically acceptable is for it to be ± 20mg/dL from the laboratory reading if glucose concentration ≤ 100 mg/dL and ± 20% from laboratory reading if the concentration is ≥ 100 mg/dL[38]. Bland-Altman analysis conducted on our data shows that most of the readings on the fabric-based sensor were within ± 10% of the laboratory value, except at the lower concentration of 78 and 80 mg/dL (Fig. 4b). This indicates that the fabric based sensors show clinically acceptable performance.

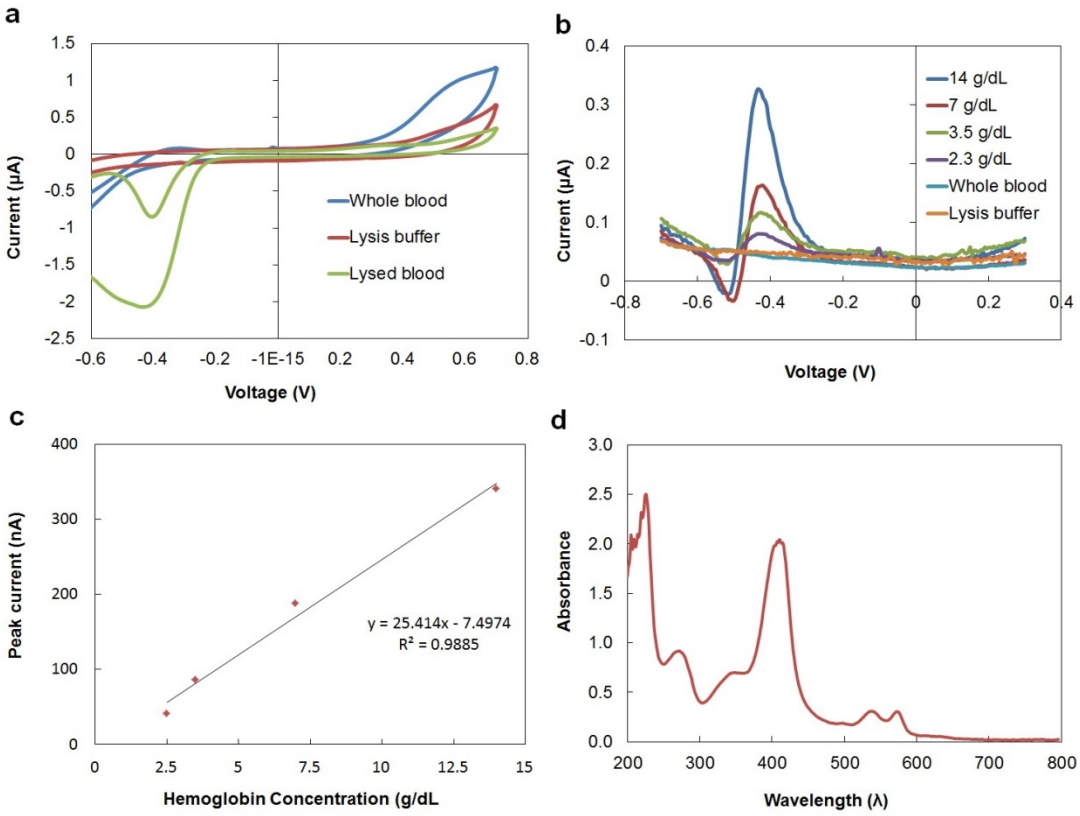

Fig. 5 Hemoglobin detection on fabric sensors. (a) Cyclic voltammetry curves of lysed blood, whole blood and lysis buffer at scan rate of 50 mV/s. Irreversible peak found at around -0.42V correspond to the peak of oxy hemoglobin. Neither whole blood nor lysis buffer showed any reduction peak. **(b)** Quantification of different concentration of hemoglobin using peak current of corresponding DPV curves at scan rate of 50mV /s and pulse amplitude of 5 mV. Increasing concentrations of hemoglobin showed increasing peak current. Each curve has a separate baseline but they are all plotted on one graph for visual convenience. (c) The exact peak currents obtained automatically from the potentiostat (Supplementary Information S1) are used to generate the calibration curve. The curve shows good linearity. **(d)** Absorbance spectra of 10X diluted lysed blood sample showing peaks at 414nm, 542nm and 577 nm. The presence of peaks at 542 and 577 indicates the dominant presence of oxy-hemoglobin.

Hemoglobin Detection

Electrochemical methods for the detection of hemoglobin have been less commonly reported when compared with absorbance and pulse oximetry. Electrochemical detection is challenging because of the extended three dimensional structure of hemoglobin[39], which inhibits the electron transfer between the deeply concealed heme group and the electrode surface. Hemoglobin is also buried deep inside the red blood cells. It is therefore necessary to lyse blood to release hemoglobin into the sample before detection. In the recent past, there have been some reports on electrochemical detection of haemoglobin using various reduced graphene surfaces and it was found that glassy carbon electrodes perform well for this purpose [40, 41]. Detection of hemoglobin on a bare glassy carbon electrode was demonstrated and an irreversible electrochemical reduction peak was observed in cyclic voltammetry and explained as the reduction peak of oxy haemoglobin and not the heme iron (Fe^{2+}) in hemoglobin[40]. In another paper, cathodic peak current of oxyhemoglobin was found at -0.410V with respect to a Ag/AgCl reference electrode and the oxyhemoglobin peak was also found to be pH and electrode material dependent[42].

In this work, we have used carbon electrodes on yarn as the working electrode to detect oxy-hemoglobin which is the dominant form of hemoglobin usually comprising 97-100% of all hemoglobin[43]. Also, when hemoglobin is exposed to ambient conditions, it normally exists as oxy hemoglobin (HbO_2) because of its special physiological function of binding with oxygen[35, 40, 41]. The presence of hemoglobin in the oxy-hemoglobin form was first confirmed by spectrophotometry (Fig. 5d). Strong peaks at 414nm, 542nm and 577nm are observed, supporting the dominant presence of oxyhemoglobin in lysed blood[39]. . In the case of dominant presence of met haemoglobin in sample 542 nm and 577 nm peak would not be observed and the peak at 414 nm would be shifted to 403 nm.

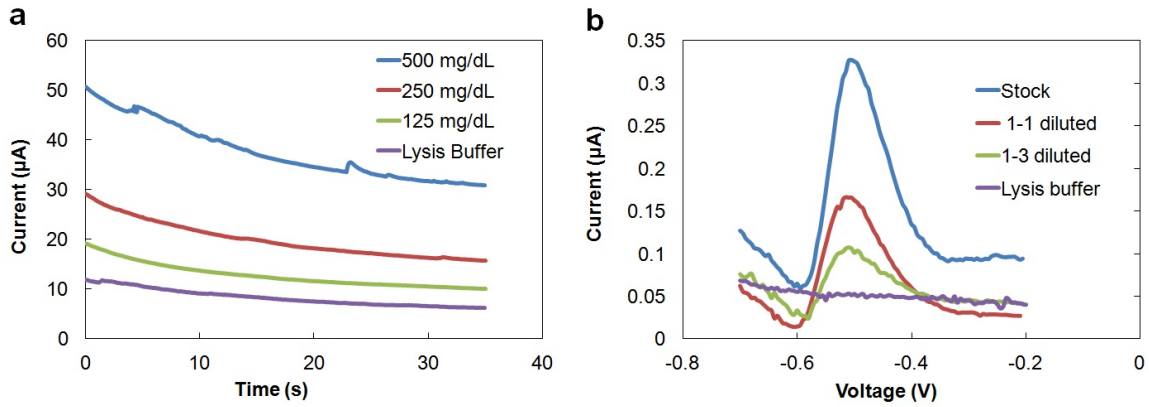

Fig. 6 Multiplexed detection of glucose and hemoglobin. (a) Detection of glucose using amperometry technique. Increasing concentrations of glucose showed higher current values at the same time point. Amperometric current decayed as the glucose in the sample was consumed. (b) Detection of hemoglobin using DPV technique after detection of glucose on multiplexed sensor (Fig. 2e). Lysis buffer (purple) did not exhibit a reduction peak at while increasing concentrations of hemoglobin showed increasing peak current.

Cyclic voltammetry was first performed to study the nature of peak with blood, lysed blood sample and lysis buffer (Fig. 5a). An irreversible peak at around -0.42 V was found when lysed blood sample was tested on the fabric sensor and this peak was absent in the other two samples. Further, no anodic peak was seen indicating the lack of any discernible oxidation reactions. We hypothesize that this peak corresponds to the reduction of oxy-hemoglobin[40] (Eqn. 2).

$$HbO_2 + 2H^+ + 2e \rightarrow Hb + H_2O_2 \quad \text{(Eqn. 2)}$$

Cyclic voltammetry is highly affected by capacitive current, which makes it very difficult to perform accurate quantification. On the other hand, differential pulse voltammetry is less likely to get affected by capacitive current as it measures the difference between the current at the beginning and end of each pulse[44, 45]. Different concentrations of hemoglobin created by diluting a stock solution were evaluated using DPV. We found that the reduction peak size measured at -0.42V progressively increased as the concentration of hemoglobin increased (Fig. 5b). The height of the peak can be used as a means to quantify the level of hemoglobin. The exact peak current was automatically generated by the potentiostat (Supplementary Information S1). A good correlation was found between the concentration of haemoglobin and corresponding peak current (Fig.5c). The minimum concentration that we were able to detect was 2.3g/dL and the maximum was 14 g/dL. Since the physiologically relevant concentration of hemoglobin is 11g/dL to 17g/dL, we believe that these sensors can detect haemoglobin from clinical samples.

Detection of glucose and hemoglobin on multiplexed sensor design

We have also demonstrated a prototype for the multiplexed detection of glucose and hemoglobin on fabric electrochemical sensors by introducing one extra electrode. The design of the multiplexed sensor for glucose, and hemoglobin (Fig. 2e) has been described above. Lysed blood sample was spiked with glucose and diluted with distilled water in proportion of 1:0, 1:1 and 1:3 which gave glucose concentrations of 500mg/dL, 250mg/dL and 125 mg/dL respectively. Sample (5μL) was dropped on the sensor from the end furthest away from the glucose sensor WE so that any enzyme or mediator should not leach and flow towards the WE of the hemoglobin sensor. Sequential detection of glucose using amperometric technique (Fig. 6a), followed by hemoglobin detection using DPV technique (Fig 6b) was performed. Amperometric readings were taken for 40 seconds for the detection of glucose and the WE for glucose was then disconnected. Thereafter, the WE of hemoglobin was connected to the reader to perform DPV. DPV was started at 1 min after dropping the sample.

Increasing concentrations of glucose showed increasing amount of current (Fig. 6a) and increasing concentrations of hemoglobin showed increased amount of peak current (Fig.6b). We also found that the amperometric and DPV peak current values reported in the multiplexed sensor were very close to the value reported for the individual single analyte sensors. This indicates that no significant interference was created while multiplexing, showing that this platform has the potential to detect multiple analytes. We believe that manufacturing and patterning for the detection of multiple analytes can be achieved quite easily through the simple insertion of an additional working electrode. The successful demonstration of this multiplexed sensor indicates that carbon electrode can be used as the reference electrode in place of Ag/AgCl electrode which will further reduce the cost of the sensor and will possibly improve sensor stability compared to sensors made with Ag/AgCl electrodes as these electrodes can interfere with test results and are more prone to getting oxidized when exposed to atmosphere or when they come in contact with a metal surface.

Conclusion

We have shown for the first time that, textile weaving can be used to mass manufacture sensors for electrochemical analysis. Tests sensors for glucose, hemoglobin and multiplexed sensors for glucose and hemoglobin were made by coating silk yarns with conducting ink and reagents and then fabricated on a simple hand-loom. The accuracy of the glucose sensors was excellent as measured by comparing results on the fabric sensor with those obtained from laboratory readings. We are continuing to work to improve the accuracy and repeatability of the sensors on the lower end of the clinical range, which is possible with increased automation in the fabrication process. We believe, with further modification in chemical composition of sensor and proper packaging, sensor can be stable for a year or more. In addition to glucose sensors, we were also able to detect hemoglobin using the technique of DPV. We have further demonstrated proof-of-concept for the multiplexed detection of glucose and hemoglobin on the same fabric sensor using amperometric and DPV technique respectively. Beyond glucose and hemoglobin testing, a number of metabolites that are used to commonly test for kidney and liver function as well as cardiovascular diseases can be tested using electrochemistry.

One of the strong motivating factors behind the development of diagnostic tests on a fabric platform was to utilize the local, low-cost and scalable technology of textile weaving to provide common diagnostic tests at affordable prices to people in the developing world. Weaving is widely practiced around the world and a strong knowledge base and skills exist in many developing countries like India. The low capital equipment costs, easy scalability and the wide availability of skills make it especially attractive for the mass manufacture of diagnostics sensors. To demonstrate how low the costs of these sensors can be, the bill-of-materials cost for glucose sensors is shown (Table 1). This is based on ordering reagents in small quantities required for research and can be further reduced with bulk price ordering. The cost of 1000 strips comes to 19.64 USD making the per strip cost <2c which can enable sales at prices that are much more affordable in the developing world than today. Equally importantly, the potential for distributed manufacture of these sensors can address problems of supply chain and logistics of delivery, which are significant hurdles to the availability of diagnostic tests at remote locations. Introducing more automation in the process can further improve performance and reduce costs through the efficient utilization of reagents like conductive ink, enzyme and mediator. For hemoglobin sensors there are no reagents (enzyme and mediator) and the absence of the Ag/AgCl electrode makes the cost of this sensor even lower than the glucose sensor. In the case of the multiplexed sensor, the cost is significantly lower than the sum of the costs of the individual glucose and haemoglobin sensor as the multiplexed sensor will contain only one extra carbon electrode for the detection of hemoglobin.

Textile weaving also provides a number of low cost options to scale up the production of these sensors (Table 2). Assuming even a 6 hour shift on a single loom, industrial quantities of sensors (~200,000 per day) can quite easily be produced using this technique with minimal capital investment ($6,000) for a power loom. Finally, processing operations and equipment common to the textile industry like dyeing, cutting, lamination and packaging can be leveraged to obtain the repeatability, aesthetics and final form factor that is required of a medical product.

Table 1 Bill of materials for the fabric glucose sensor.

Material	Cost per unit ($)/g	Consumption of material per 1000 sensor (g)	Cost per 1000 sensor $
Carbon ink	0.75	10	7.5
Silver (Ag/AgCl) ink	2	3	6
Enzyme	0.4	0.30	0.01
Mediator	0.3	3.3	0.99
Backing card	-NA-		5
Silk and other thread cost			0.13
	Total cost of 1000 sensors		19.64

Table 2 Daily production capacity of glucose sensors on different looms

Loom type	Weft insertion Speed/minute	Loom width	Sensor Production per day(6hr)	Cost of one loom (USD)
Hand loom		2 m	20,000	1,000
Shuttle power loom	120 weft/minute	2 m	91,440	2,000-3,000
High speed power loom	200 to 300 m/min	2m	182,880	3,000-6000

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Notes

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