MATTRESS-BASED SWEAT MONITORING FOR HUMAN HEALTH MONITORING AND SMART HOMES

Spyridon Pavlidis[1,2], Meng-Yen Tsai[2], Billyde Brown[2], Decarle Jin[2], John-David Velilla[3], Michael DeFranks[3], and Eric Vogel[2†]*
[1]North Carolina State University, Raleigh, North Carolina, USA
[2]Georgia Institute of Technology, Atlanta, Georgia, USA
[3]Serta Simmons Bedding, Norcross, Georgia, USA

ABSTRACT

Mattress-based sweat sensing systems can provide non-intrusive, continuous health monitoring. Commercialization of physical sensors in beds has already been achieved, while chemical sensing within the same platform is still in its infancy. We describe our approach towards the development of flexible potentiometric sensors and demonstrate low-temperature fabrication of these devices. Three commonly used mattress fabrics have been contrasted as sweat capture layers, leading to pH sensing with a sensor embedded within the hydrophilic fabric. It is found that noise, sensitivity and linearity can be affected by the rate of liquid flow, temperature of the Al_2O_3 deposition and stability of the thin film Ag/AgCl reference electrode realized on-chip.

INTRODUCTION

As electronics continue to become smaller, the sensors market has penetrated deeper into our daily lives, ranging from personal use to industrial use, among other applications. Mainstream commercial sensors in the market today largely rely on physical sensors, such as accelerometers and gyroscopes within smartphones, to track movement. Wearable sensors of this type have also emerged, allowing for more continuous and less intrusive capture of data. Wearable chemical sensors, however, remain a very active subject of research, with recent work focused on sweat analysis to assess one's health [2-4]. Compared to blood, sweat is a particularly attractive analyte since it can be acquired non-invasively and contains chemical markers with significant physiological correlation [5]. Despite the many uses of wearable sensors, in particular when the user is in transit and activity tracking is of interest, there exist applications where wearing a device is not practical or even possible.

One such scenario is health monitoring while lying on a mattress, which can be used both for everyday sleep analysis, as well as in-patient monitoring for bed-bound patients. Since humans spend an average of one third of their lives lying on mattresses, they represent a unique yet heretofore untapped opportunity to monitor one's health and enable smart home systems that tailor our environments to optimize our wellbeing. Unlike wearable sensors, in-built mattress sensors would not require user installation or training and would therefore facilitate use. They also do not interfere with one's comfort, or force a change of habit in terms of what is worn during sleep.

In this work, we investigate the development of flexible chemical sensors embedded in mattresses for sweat-based health monitoring. To our knowledge, this is the first time that flexible chemical sensors have been integrated into mattress materials to detect pH, which is tied to electrolyte [6] and lactic acid [7] concentrations, among other constituents. This paper introduces our mattress-integrated sweat sensing approach, describes the implementation challenges and associated design choices that have been made to make progress towards this system. A low-temperature and scalable microfabrication process is also described, with prototypes demonstrated on both silicon and flexible polyimide substrates. We compare three commonly used mattress fabrics for their suitability as liquid-absorbing interfaces to the sensor's surface. By leveraging an extended gate sensor configuration, we also present results on the performance of these sensors when integrated into the mattress material, the effect of decreasing deposition temperature for the pH-sensitive dielectric thin film, and the stability of the on-chip reference electrode.

MATTRESS SENSOR CONCEPT AND DESIGN

A first generation of mattress sensors has already been commercialized by Serta Simmons Bedding as accessories to their Beautyrest® line of mattresses (Figure 1) [1]. The Sleeptracker® sensing system consists of physical sensors and a processing unit that can be installed between the mattress and the bed frame/base. The sensors are responsible for tracking the heart and breathing rates as well as the motion/movement of the bed's occupants, while the processor, which is located underneath the base/frame operates the sensors, collects the data and transfers them to the cloud for storage. The data can then be retrieved by users on their smartphone or computer. Access to this information allows the user to monitor the quality of their sleep by quantifying the time spent in different sleep cycles, without having to wear a bracelet or other device. Serta Simmons's aim is to enable users to improve the quality of sleep – for example, they can optimize the timing of a pre-programmed alarm to wake a user during the most comfortable sleep cycle.

This infrastructure forms a strong foundation for the development of a new generation of chemical sensors, which is the focus of this work. However, the design of sweat-sensitive sensors must address new challenges compared to the already existing physical sensors. Design requirements include: close proximity to the sleeper in order to collect the sweat prior to sample spoiling or evaporation; operation with low sample volume; high sensitivity;

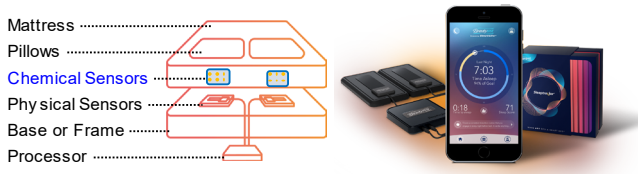

Figure 1: Serta-Simmons Bedding has commercialized a Beautyrest® line of mattresses with Sleeptracker® technology that does not necessitate the use of wearable devices. (a) These physical sensors are placed in between the mattress and bed base/frame to monitor one's heart and breathing rates, as well as their movement to report the time spent sleeping. The focus of this work is the development of novel chemical sensors, located in the upper layers of the mattress, to detect sweat content. (b) The sensor data is collected by a processor unit, which uploads information to the cloud and can be accessed remotely. [1]

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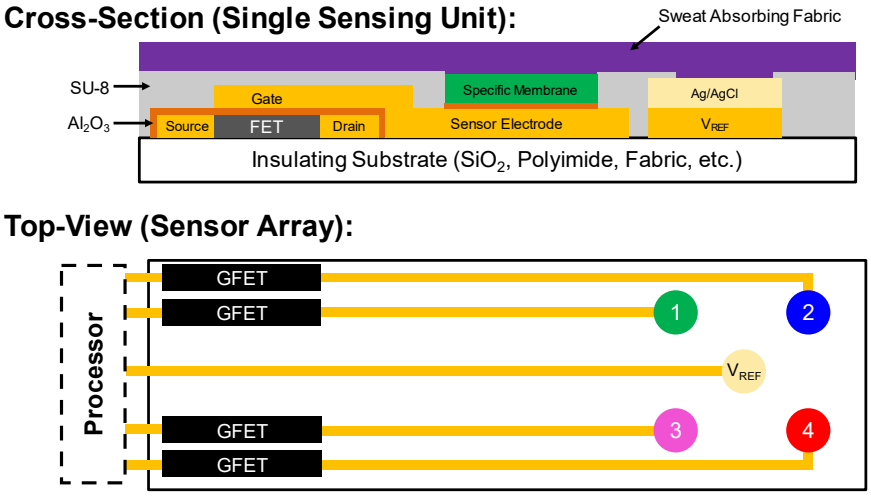

Figure 2: System overview for mattress-integrated sensor array for sweat analysis to provide health monitoring and smart home capabilities. The cross-sectional view shows the critical components of a single module in the system, including the functionalized sensor electrodes, while the top-view demonstrates how each module can be cascaded to realize the multi-component sensing array. The FETs are depicted conceptually as potentiometers.

high selectivity to detect different components of the sweat; and a thin, flexible form factor so as to not interfere with sleep quality. Additional system-level challenges include data transfer, power supply and long-term stability or ease of replacement. The latter two are of particular importance since the average lifetime of a mattress is ten years.

An illustration of the conceived sweat-sensing system for in-mattress integration is depicted in Figure 2. The cross-section outlines the three major components of each sensor: (1) sensor electrode, (2) reference electrode and (3) FET transducer. The entire platform is fabricated on an insulating substrate. The sensor electrode consists of an alumina (Al_2O_3) thin film deposited on gold. Al_2O_3 has been chosen since it can be deposited with high quality at a wide range of temperatures and has been demonstrated to be pH sensitive [8]. To enhance specificity, an analyte-specific layer can be added on top of the Al_2O_3, which will dictate that any change in potential observed at the Al_2O_3's surface is due to the change in concentration of the specific ion in question. In close proximity to the sensor electrode, a Ag/AgCl reference electrode is formed. This electrode is critical to this work, and it is important to characterize and consequently improve its operational stability.

The sensor electrode is connected to the gate electrode of a charge-sensitive field-effect transistor (FET). This separation of the FET and sensor surface is known as the extended gate configuration, and serves two advantages compared to the traditional ion-sensitive field-effect transistor (ISFET) layout: (1) it enhances operational reliability of the FET by isolating it from direct contact with the liquid-phase analyte, and (2) permits independent development of the sensor electrode from the FET device itself. Graphene has been identified as a suitable channel material for the FET since it can be deposited over large areas and has already been demonstrated for use in potentiometric biosensors [9]. The FET and metal traces on-chip are protected by an SU-8 microfluidic isolation layer. This thin film is patterned with openings located only at the sensor and reference electrode surfaces. Given the aim of this work to integrate the sensors into a mattress, the top of the chip is fitted with a mattress fabric. This material must be carefully chosen to absorb sweat and consequently bring it into contact with the sensor's surface.

The bottom half of Figure 2 presents a top-side view of a sensor array that can be implemented using the architecture described above. To minimize size requirements, a common reference electrode is leveraged for multiple, individually-functionalized sensor electrodes to detect different components of sweat. Each of these is connected to a potentiometer (i.e., the high impedance gate of a FET) whose respective signals are fed to a computer or microcontroller that processes the information and makes it available to the user for analysis.

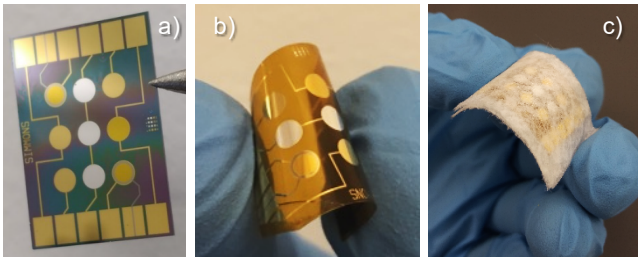

Figure 3: (a) Extended gate sensors fabricated on silicon substrate, (b) flexible sensors fabricated on polyimide substrates and (c) packaged in fabric used in commercial mattress.

SENSOR FABRICATION

The sensors were fabricated on both silicon (Figure 3a) and polyimide (Figure 3b) substrates. In this paper, we focus on the development and characterization of the sensor and reference electrodes, paying particular attention to the effects of low-temperature fabrication on their performance and methods to improve them. We therefore employ a FET-less fabrication flow, made possible by the extended gate configuration described above. By using flexible substrates, direct embedding of the sensors within a mattress fabric can be achieved, as shown in Figure 3c. In the case where silicon (Si) wafers were used, a thermal silicon oxide (SiO_2) was created prior to metal deposition. To improve metal adhesion as well as robustness to wet processing when working with the polyimide substrates [10], they were first annealed in vacuum at 200 ˚C for 8 hours, treated with O_2 plasma for 60 sec and then both sides were coated with 150 nm of plasma-enhanced chemical vapor deposited PECVD silicon nitride (SiN_x) at 100 ˚C. A first metal layer consisting of chrome/gold/aluminum (Cr/Au/Al) with thickness 25/300/2 nm was deposited using electron-beam evaporation and patterned using lift-off. The top Al surface was descummed using an O_2 plasma that also facilitates the adhesion of the consequent atomic layer deposition (ALD) of Al_2O_3. The Al_2O_3 layer was deposited at 250 ˚C and 150 ˚C with a thickness of 15 nm, the effects of which are discussed below. In order to gain access to the metal layers, vias are etched into the Al_2O_3. This is executed using Al Etchant (based on phosphoric acid) that is heated to 50 ˚C for 2.5 minutes. The samples are once again descummed with O_2 plasma, prior to the electron-beam evaporation of the reference

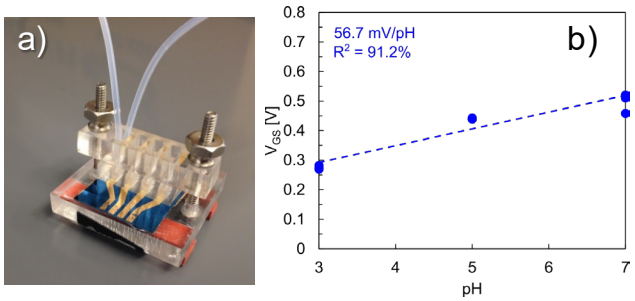

Figure 4: (a) Initial extended-gate samples were fabricated on silicon wafers using shadow masking techniques to produce high-quality interfaces. For testing, they were packaged in PDMS microfluidics and off-chip FET was used as a transducer. (b) For a reference current of 100 nA, a sensitivity of 56.7 mV/pH was observed, which is near-Nernstian.

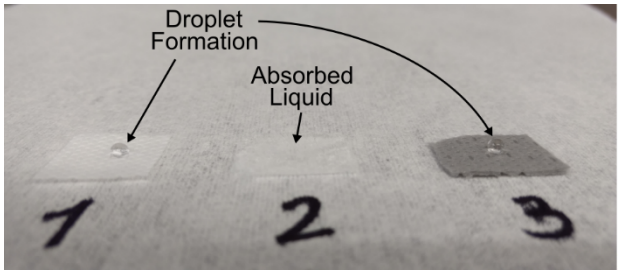

Figure 5: In order for analyte to reach the sensor's surface, the mattress fabric must be hydrophilic. Three materials commonly used by the mattress industry were tested: 1) spun bonded, nonwoven polypropylene, 2) long staple fiber, needle punched, nonwoven polypropylene, 3) spun bonded and needle punched, nonwoven polypropylene. With Samples 1 and 3 exhibiting hydrophobic behavior, Sample 2 was chosen.

electrodes consisting of titanium/silver (Ti/Ag) thin films with 30/150 nm thickness, which are also patterned using lift-off. SU8-3005 is then spin-coated and patterned, resulting in a 5 µm thick microfluidic isolation layer. Finally, the exposed Ag thin film is chloridized to Ag/AgCl by drop coating ferric chloride ($FeCl_3$)-based copper etchant for 1 minute, before rinsing thoroughly with DI water.

RESULTS

To establish a baseline of performance against which to compare the microfabricated sensors, a set of chips was fabricated using shadow masks and tested using PDMS-based microfluidics, as shown in Figure 4a. Since initial process development was performed on Si substrates without temperature constraints, the Al_2O_3 layer was deposited at 250 ˚C. The sensor electrodes were connected to the gate of an FET, while the on-chip Ag/AgCl reference electrodes were biased in order to capture V_{GS}-I_D for each pH solution. It is known that a change in the pH on the dielectric surface results in an effective change in the threshold voltage (V_{TH}), which can be extracted from the FET's transfer curve. The maximum sensitivity is known as the Nernstian Limit, which corresponds to 59 mV/pH. To establish some stastical significance, each measurement was completed four times. Figure4b shows the change in V_{GS} for a reference current of 100 nA. From this, it is gleaned that the sensitivity is 56.7 mV/pH (very close to Nernstian) and the linearity is 91.2%.

Three samples of mattress material were evaluated for their suitability of integration with the sensor. This test was conducted by dropping water droplets onto the surface and consequently

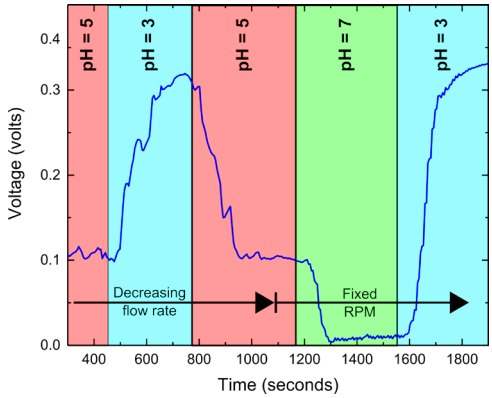

Figure 6: The potentiometric transient response of the sensors to various pH solutions was measured. The backside of the chip was attached to flexible PDMS and embedded within the liquid absorbing mattress fabric, the latter now replacing the conventional microfluidic structure.

determining if they were absorbed by the fabric, or retained as a droplet. The three materials that were compared were: 1) spun bonded, nonwoven polypropylene fabric, 2) long staple fiber, needle punched, nonwoven polypropylene fabric, 3) spun bonded and needle punched, nonwoven polypropylene fabric. As shown in Figure 5, Sample 2 is the only one that absorbed the droplet and will therefore serve as a good top interface for the sweat sensor

To address mattress integration, low-temperature sensor chips were fabricated with a maximum thin film deposition temperature of 150 ˚C in order to be compatible with the thermal budget of polyimide and other flexible substrates. These were first characterized on a silicon-based chip whose backside was attached to soft and flexible PDMS for support while the top was covered with the hydrophilic fabric identified above. Using a peristaltic pump, it was possible to control the flow rate of pH solutions that were dropped onto the fabric to study the effect of flow rate, as well as switch between different pH values to assess sensitivity. Transient data is depicted in Figure 6. From t = 0 to t = 1100 sec, the flow rate was decreased from 40 RPM to 3 RPM. During this window, there is observable noise in the voltage response. After t = 1100, where the flow rate remains 3 RPM, the noise decreases significantly. The relationship between signal noise and flow rate necessitates further study, and the consideration of design techniques to reduce its effect. Control of the flow rate could be achieved through localized mattress heating and/or iontophoresis. The sensitivity of the response is also non-linear. For the transition between pH = 5 and pH = 7, the sensitivity is 48.2 mV/pH. This is slightly less than what was observed when high temperature Al_2O_3 was used in Figure 4b, but still sufficient. Moreover, the sensitivity for the transition between pH = 7 and pH = 3 increased. The source of this non-linearity was investigated by examining two variables: (1) the effect of the ALD Al_2O_3 temperature, (2) the on-chip Ag/AgCl reference electrode's stability in liquid.

To isolate the impact of ALD temperature on pH sensing performance, sensors with Al_2O_3 deposited at 100 ˚C were tested in PDMS-based microfluidic structures, with I-V curves extracted from a gate-connected FET and an off-chip flow-through Ag/AgCl reference electrode. As shown in Figure 7, the sensitivity of these films was found to be 23.8 mV/pH with a significantly decreased linearity of 66.6%. It has been noted in the literature that oxidation of Al_2O_3 films can improve pH sensing performance [11]. Thus, the sensors were subjected to 30 sec O_2 plasma at 50 W before another round of testing. This treatment resulted in an improvement in both the sensitivity (34 mV/pH) and the linearity (74.5%). Further

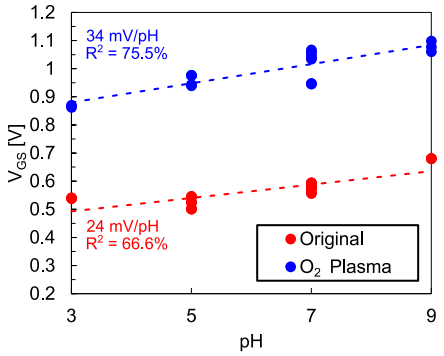

Figure 7: It was found that reduction of the ALD Al_2O_3 deposition temperature to 100˚C lowered both the sensitivity and linearity of the pH response compared to the original 250˚C case. This can be counteracted through the use of O_2 plasma treatment on the film.

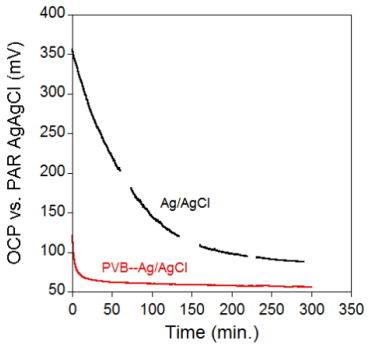

Figure 8: The stability of the e-beam evaporated Ag/AgCl reference electrode can also contribute to the response, but can be improved through the use of a PVB coating.

investigation is required to understand this effect, though it is hypothesized at this stage that it could be related to the impact of film stoichiometry, with decreased oxygen incorporation in Al_2O_3 when deposited at low temperature requiring compensation via O_2 plasma versus direct high temperature deposition.

Most pH sensors in the literature rely on off-the-shelf macro-scale Ag/AgCl reference electrodes for demonstration. In benchtop testing scenarios, this is an acceptable approach, however it cannot be adopted in our application, which requires full on-chip integration. It is therefore important to characterize the effect of the thin film-based reference electrode we have constructed. Figure 8 demonstrates that over a period of 5 hours, a bare Ag/AgCl reference electrode subjected to constant bias in a KCl solution can exhibit over 200 mV of drift. To combat this, previous studies [4, 12] have proposed the use of a polyvinyl butyral (PVB) coating. We have tested this approach and found that it indeed reduces the "burn-in" time required for the electrode to reach a stable state. While other studies have examined these reference electrode preparation techniques for short- to medium-term use (e.g., disposable wearables), further study is required to understand their long-term stability, for example over one year. Care must also be taken to characterize the reference electrode's potential versus pH and ionic strength to ensure stability against these variations.

CONCLUSION

Wearable sensors offer much potential in terms of continuous health monitoring, but there exist scenarios where sensors fully integrated into the objects of our environment can help improve quality of life. In this work, we have presented an approach to embedding potentiometric sensors into mattress fabrics for sweat analysis during sleep. We have described an architecture for such a system, and demonstrated microfabrication of these sensors. Our results reveal several factors impacting the performance of these sensors as we move away from conventional microfluidic approaches, such as the rate of analyte capture, the low temperature deposition of the pH sensing Al_2O_3 film and the stability of the on-chip Ag/AgCl reference electrode. Methods to successfully combat these effects have been presented, thus motivating further study and development of these sensors.

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CONTACT

*S. Pavlidis, tel: +1-919-513-3018; spavlidis@ncsu.edu
†E.M. Vogel, tel: +1-385-7235; eric.vogel@mse.gatech.edu