Monitoring of Respiration by Means of an Additively Manufactured Barium Titanate-based Hygroscopic Sensor

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***Abstract*—Unobtrusive respiration monitoring is critical for a number of applications such as monitoring of sleep apnea and changes in respiration rate. Wearable, flexible sensors provide increased comfort and versatility over traditional rigid sensors. Recent advances in additive manufacturing, such as printed electronics, allow miniaturization of sensors and implementation onto flexible, unobtrusive substrates. In this paper we present implementation and validation of a printed co-doped barium titanate capacitive sensor that is highly sensitive to water vapor as an unobtrusive breathing sensor. This attribute allows the sensor to identify inhalation/exhalation cycles, detect apnea conditions and hyperventilation, and report respiration rate. The 2x4mm sensor demonstrated capacitance changes of up to 1000% during breathing, while placed directly under the nostrils. In addition, we demonstrate that the sensor increases capacitance from 175pF in ambient conditions up to 225pF (almost 30%) during breathing at a distance of 30cm while sampled at 200Hz. We demonstrated viability of this printed sensor for both wearable monitoring and ambient sensing applications.**

Keywords—printed sensor, hygroscopic, wearable sensing, respiration monitoring, breathing

I. Introduction

Continuous respiratory monitoring enables the early detection of diseases and disorders such as sleep apnea and cardiac arrest and can also help with the characterization of illnesses like COPD and asthma. Respiration rate is a vital measurement (akin to heart rate, blood pressure, body temperature etc.) that is essential for medical professionals to assess the overall wellness of a patient; yet it is the most underreported metric because reliable medical devices for continuous monitoring are either impractical or unavailable. There exists a real need for noninvasive devices that are capable of continuous monitoring, able to reliably alarm if breathing becomes too slow, ragged, stops, etc., and be easily set up to monitor by a medical professional. Many approaches have been taken to measure respiration rate by sensing air flow [2], [3], [4], [5], measuring abdomen and thoracic circumference [6], or body impedance [7]. This paper will investigate an approach that uses a low-cost capacitive sensor to measure air flow. Measurements will be taken by connecting to an embedded platform to obtain high sensitivity respiration data and calculate respiration rate. This sensor uses a printed ceramic material as a dielectric which was developed by NASA Marshall Space Flight Center [1]. It exhibits a highly sensitive response to changes in moisture with very fast response times (<1second).

II. Survey

A. Airflow Sensing

The method of airflow sensing as a way to measure respiration rate involves placing a device directly under the nose or mouth of the patient to monitor the air being exhaled during respiration [2], [3], [4], [5]. These sensors can be temperature, humidity, CO_2, or sound detection sensors that sense the increase in relevant stimuli that result from air being exhaled onto them. Due to the nature of direct sensing, these sensors often need to be worn and are affixed to a mask or use a device to mount just beneath the nose. Some researchers have made these wearable sensors to communicate wirelessly to report data, so the device can be worn more comfortably without trailing power and communication cords.

Recently, *Caccami et al.* [2] used a wireless graphene-oxide based hygrometer affixed to a mask and connected to a Radio Frequency Identification (RFID) chip to sense respiration rate and wirelessly monitor it. With this device, the researchers were able to experimentally demonstrate the detection of inhalation/exhalation and identify abnormal breathing events like apnea. This was done by obtaining the resistance of the sensor through the RFID sensor and correlating the variations to inhalation and exhalation. The benefit of this design is that the sensor system is passive and does not require a battery at the user end. However, it has a limited communication range, requires sophisticated RFID interrogation equipment, and the sensor appeared to saturate after several hours use.

A humidity sensing approach was also taken in *Kano et al.* [3] where a resistive SiO_2 sensor was connected to a headset and signal processing unit that could be worn during activities such as exercise. This device collected data and communicated wirelessly via Bluetooth allowing a much greater range than the RFID approach. The addition of signal processing to the unit allowed much higher fidelity of the data obtained and the researchers identified many different respiratory events with the single sensor unit.

Gupta et al. [4] also used the sensor-in-mask approach but used thermistors to detect changes in temperature and correlated that to respiration. The idea was to create an extremely low-cost device that could be implemented as a standalone in low-resource areas. The main functionality required for this application was a read out of respiration rate and alarms triggered by respiratory distress events. This is a less sophisticated system without any wireless communication but accomplishes its goal of low-cost monitoring.

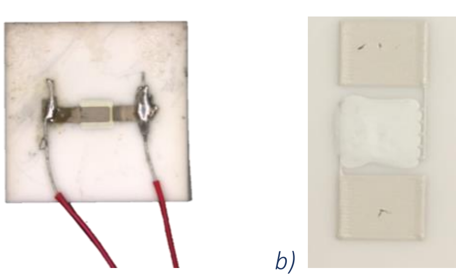
Mohammadhi-Koushki et al. [5] created a hybrid wearable device that was capable of monitoring heart rate and respiratory rate by sensing the sounds associated with both. This device communicated wirelessly with a smart phone for data processing making it another portable solution to respiration rate monitoring.

B. *Sensing changes in movement and volume*

Several sensors have been developed to monitor respiration rate by detecting the movements of the body that are associated with respiration. These approaches have used magnetometers to sense slight perturbations while sleeping, devices mounted on the chest to measure the expansion of the chest during inhalation, and electrodes to observe changes in impedance during respiration of the thoracic cavity. Many of these methods are very precise but require rigid circumstances to get accurate measurements.

Merritt et al. [6] created a textile based capacitive sensor that was meant to be worn as a belt around the chest to monitor respiration rate as a change in capacitance when the fabric expands. This system had micrometer precision and was able to detect circumferential variation up to 60 mm.

A magnetometer design was used in *Milici et al.* [7] to characterize the apneas of patients during sleep studies. This sensor was printed onto a miniature board which included Bluetooth Low Energy (BLE) communication. The microprocessor-based signal processing system was able to quantify respiration rate, apnea duration, and movement time.



a) b) Fig 1. Sensor implementation: a) screen printed parallel plate capacititve sensor, b) micro-dispensed interdigitated capacitive sensor.

III. Barium Titanate-based Sensor

The sensor reported in this study is made with a co-doped barium titanate (BaTiO3) dielectric and palladium-silver (Pd-Ag) electrodes via an additive manufacturing process. For this project, both a high temperature processed screen-printed construction and a low temperature processed micro-dispensed construction were evaluated.

A. *Screen-Printed Sensor*

This sensor (Figure 1a) is made using a layered screen-printed approach with inks that cure at high temperatures to ensure the sensors can withstand some physical stress and harsh environments. The sensor consists of three layers printed onto an alumina substrate in the following configuration: 1) Pd-Ag base electrode 2) Co-doped BaTiO3 dielectric with a lead germanate glass binder and 3) Pd-Ag top electrode to create a parallel plate capacitor. Each layer is individually cured at 850°C to sinter the ink layer. The resulting sensor has dimensions of 2x4mm with approximately 50µm dielectric thickness which functions as the sensitive area. At a relative humidity of 40% the sensors measure between 100-200pF with response up to 1-10nF at 90% RH.

B. *Micro-Dispensed Sensor*

This sensor (Figure 1b) is made using an nScrypt n300 machine to micro dispense low temperature curing inks onto a variety of substrates. This method adds the versatility of additive manufacturing, which lends itself to complex geometries and the use of many different substrates including FR-4 and Kapton for direct PCB integration and flexible sensors, respectively. This sensor is constructed using interdigitated silver electrodes with a bulk Co-doped BaTiO3[1] dielectric printed on top. Both of these layers are cured individually at 120ºC. At a relative humidity of 40% the micro-dispensed sensors measure between 10-20nF with response up to 100µF at 90%RH.

C. *Sensor Selection*

Both sensor types are suitable for this application as respiration monitoring does not subject the sensors to extreme environments or physical stress. For wearable applications, the micro-dispensed sensor could be preferable as a flexible substrate could be selected. However, due to the small footprint (2x4mm) of the screen-printed sensor, its rigid substrate does not preclude its use in wearable devices. Therefore, the deciding factor for sensor selection was the dynamic range of capacitive response of the two sensors. The Teensy 3.2 microcontroller platform’s TouchRead function is used to perform a Capacitance-to-Digital (CDC) measurement on the sensor. This function has a maximum measurement value of 1300pF and a 16-bit resolution. Due to this range and pending a redesign of the micro-dispensed sensors to lower capacitance, the screen-printed sensors were chosen for experimentation and prototyping.

IV. SYSTEM DESIGN

As mentioned above, a Teensy 3.2 microcontroller was used to measure the sensor's capacitance during experimentation. A negative thermal constant thermistor was also connected to the Teensy board to observe changes in temperature as a result of breathing for verification of the printed sensor. The printed sensor, thermistor, and the Teensy micro controller were affixed to a nasal cannula. This ensured proper placement of the thermistor which was determined necessary to be into and centered in the nostril to observe sufficient temperature changes to gather meaningful data. The printed sensor was attached around 30 cm away from the nostrils demonstrate the small amount of air flow necessary to measure moisture changes. The Teensy microcontroller was connected via a serial connection to a computer to monitor the data in real time and record it for later processing.

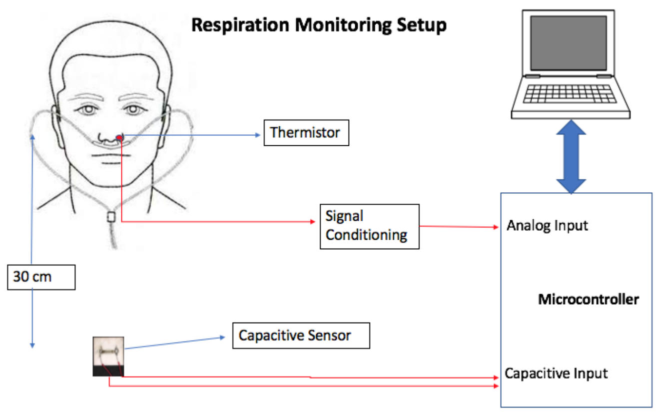


Figure 2 Experimental setup with thermistor connected to the cannula and capacitive sensor placed 30cm away

V. RESPIRATION MONITORING EXPERIMENTS

A. Validation with Known Sensor

The first experiment measured the responses of the printed sensor and thermistor simultaneously while the test subject breathed normally. The data was used to verify that both sensors were operational and sensed new exhalations at the same time. As shown in Fig. 3, the two sensors agree perfectly in observing new exhalations.. Furthermore, it can be seen that the capacitive sensor shows a greater response as a percentage with less noise than the thermistor despite being placed in an area with much less air flow. Use of a Finite Impulse Response (FIR) filter on the thermistor data was necessary due to the high level of noise in the sensor. This plot shows how the two signals align. This test verified the functionality of the printed sensors and demonstrated that the printed sensor data does not require any filtering to easily observe respiration patterns.

B. Testing Respiration Events

Once the sensor was proven to be functional, the response of the sensor was tested to measure its response under a variety of respiration events. These include normal breathing,

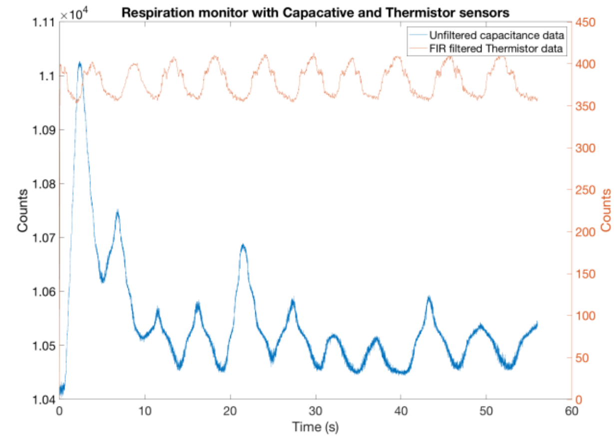


Fig 3. Filtered thermistor data in red plotted with raw printed sensor data in blue

hyperventilation, normal breathing with hyperventilation events, and normal breathing with apnea events (as shown in Fig 4 a-d).

The normal breathing test shows that the sensor increases and decreases capacitance fast enough to return to baseline before the next breath. This avoids the baseline creep many other sensors experience and results in a large difference from inhale to exhale (peak to valley) that stays consistent. It can be observed from the extreme hyperventilation test that each breath is still discernible, but the baseline does increase because the sensor does not have time to recover before the next breath occurs. For respiration rate measurements this should not be problematic however, as the peaks are defined and the time in between them easily calculable. The apnea test shows that the apnea condition is easily resolvable and that the baseline noise of the sensor is low enough that "false breaths" are unlikely. Furthermore, the sensor returns to normal behavior once the apnea condition ends. These characteristics are important because a system must detect and send an alarm if apnea occurs. Not sending an alarm when necessary is much more dangerous than a false alarm for patients. Similar to the behavior during apnea, the breathing with hyperventilation test shows that the sensor can respond quickly, and the sensor returns to normal operation after the hyperventilation ends.

C. Two Sensor Implementation

To demonstrate versatility and expandability, a second printed sensor was added to the system. The cannula was modified to direct flow from each nostril over each sensor separately. This implementation allowed for the separate sensing of the flow from each nostril. Results from this test are shown in Fig 5.

D. Calculating Respiration Rate

In addition to output of raw sensor data by the system, physiological monitoring applications usually require real-time assessment of the respiration rate.

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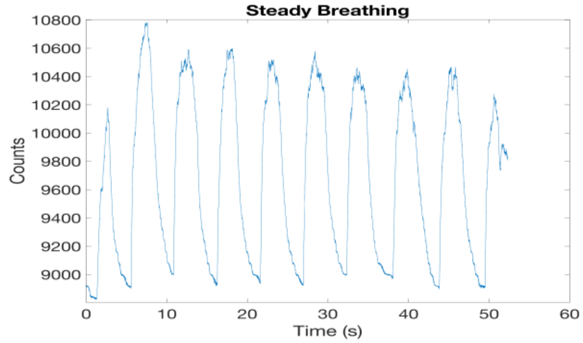

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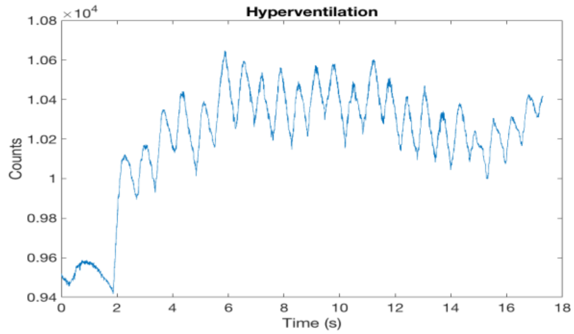

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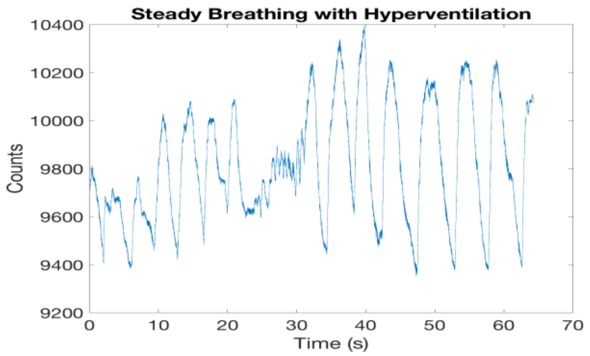

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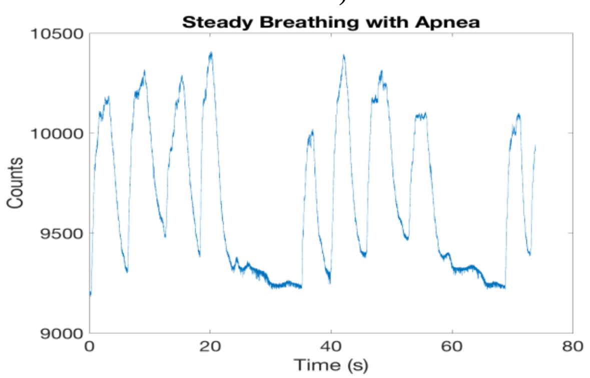

Figure 4. a) Steady breathing, b) hyperventilation, c) steady breathing with hyperventilation events, and d) steady breathing with apnea events

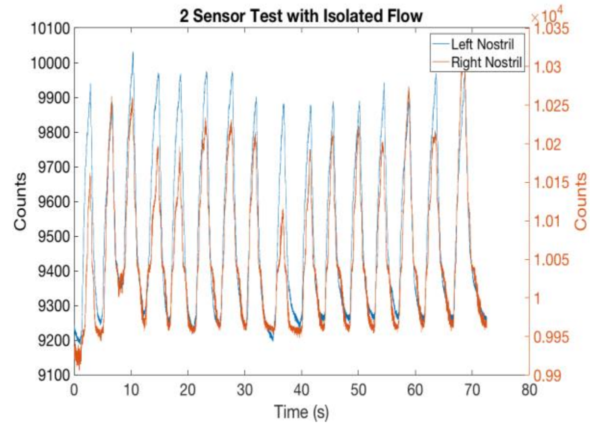

Figure 5. Two-sensor implementation with air flow from each nostril isolated

This was done so that in a final implementation of the system, where wireless communication replaces the wired serial connection, low power modes are possible. In low power mode, data transmission will only occur when a new respiration event is detected. This will extend the battery life of the device.

Respiration rate was calculated on the Teensy processor by calculating peaks in real time. The algorithm compares the most recently read value, to the last set maximum and determines if the new value is a maximum. This continues until no new maximum has been set for a predetermined period of time. The system then calculates the time between the current maximum and the last maximum, calculates a respiration rate as breaths per minute, and transmits this rate. Transmissions are sent as weighted pulses. These weights are proportional to the time between peaks. The receiving side reads these weights or times the pulses to display respiration rate. This algorithm counteracts the effects of noise in the sensor data so that true maximums are reported and the calculated time between peaks is accurate. Breathing events with corresponding weighted pulses are shown in Fig 6.

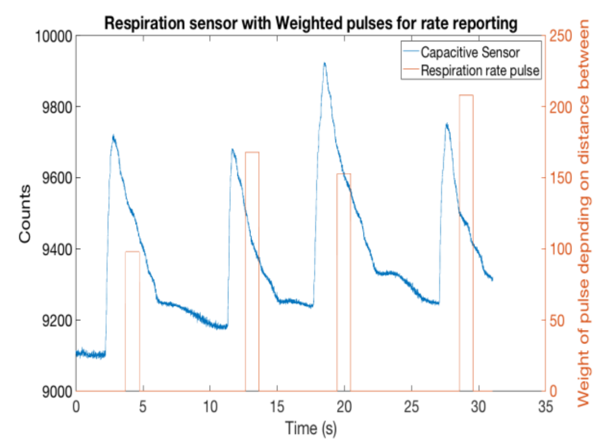

Figure 6. Breathing events in blue with weighted pulses in orange denoting detected respiration events.

VI. Discussion and Conclusion

Unobtrusive monitoring is essential for acceptance and adoption of the wearable monitoring applications. Wearable, flexible sensors provide increased comfort and versatility over traditional rigid sensors that are bulky and large. Through this work, the efficacy of the NASA developed dielectric material as a respiration-sensing device was presented. The response to breathing-induced humidity changes of the devices were shown and characterized into meaningful respiration data. The system that was designed and implemented in this study has many advantages over currently used systems. Mainly, the reported system is inexpensive (each sensor costs ~$.20 to manufacture) and the sensors used are more sensitive than the other direct flow sensors cited within. We demonstrated these sensors are capable of exhibiting 30% change of capacitance due to breathing at distances of 30cm, such a working distance enables ambient monitoring of breathing. Breathing sensors can be placed away from the nose as opposed to direct nostril placement, which increases patient comfort without compromising monitoring capabilities. Future developments of this system include wireless communication between the sensor and a home server or a smart phone.

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