Long Term Monitoring of Respiration and CO2 using Flexible Printed Sensors

Emil Jovanov	Ian K. Small	Terry D. Rolin	Harsha Ganegoda	Curtis Hill
The Univ of Alabama in Huntsville	NASA Marshall Space Flight Center	NASA Marshall Space Flight Center	The Univ of Alabama in Huntsville	NASA Marshall Space Flight Center
Huntsville, AL 35899	Huntsville, AL	Huntsville, AL	Huntsville, AL 35899	Huntsville, AL
emil.jovanov@uah.edu	ian.k.small@nasa.gov	terry.d.rolin@nasa.gov	hb0038@uah.edu	curtis.w.hill@nasa.gov

Abstract— **Long term unobtrusive monitoring of respiration and environmental conditions is critical for tracking the health and psychophysiological state of astronauts. Wearable, flexible sensors provide increased comfort and improved human factors. Recent advances in additive manufacturing, such as printed electronics, allow miniaturization of sensors and implementation onto flexible, unobtrusive substrates. We developed a low power, flexible, sensor platform with wireless Bluetooth Low Energy (BLE) interface. The platform integrates 9 Degrees of Freedom (DOF) inertial sensor, ambient sensor (temperature, humidity, and Carbon Dioxide/Volatile Organic Compounds - CO2/VOC sensor), and prototyping area for printed sensors. In this paper we present an implementation of a printed co-doped barium titanate capacitive sensor that is highly sensitive to water vapor as an unobtrusive breathing sensor. The 2x4mm sensor is highly sensitive to changes during breathing, even at distances of more than 20cm. Synergy of information from on-platform and printed sensors allows long term user monitoring both as wearable and ambient sensor. We demonstrate the results of sensor validation using 9 subjects. Average change of capacitance of the sensor during breathing at the distance of 7.5 cm was 6.2±3.5 pF. Our sensors represent very good option for low-power, wearable, and unobtrusive monitoring.**

TABLE OF CONTENTS

1. INTRODUCTION

Unobtrusive monitoring of respiration is very important for space applications. Respiration involves complex interaction of several subsystems. It is a unique process that is both voluntary and involuntary and represents physiological and psychological state of the subject. Traditionally, medical evaluations use respiration rate (RR) that is mostly counted manually as the number of breaths per minute (BPM).

Changes in respiratory rate can indicate cardiopulmonary arrest, chronic heart failure, pneumonia, and other conditions. RR is a better discriminator than blood pressure and pulse rate in identifying high-risk patient groups of cardio-pulmonary catastrophic deterioration [1]. Continuous monitoring during long periods may facilitate the early detection of diseases and disorders such as sleep apnea and cardiac arrest, and monitoring of chronic diseases, such as Chronic Obstructive Pulmonary Disease (COPD) and asthma. Respiration rate is a vital measurement that is essential for medical professionals to assess the overall wellness of a patient; however, reliable unobtrusive monitoring necessary for long-term continuous monitoring is still not available. Wearable and ambient monitoring, as described in Section 2, provides promise for continuous monitoring of respiration and support for health, wellness, and behavioral monitoring applications.

This paper presents preliminary results in the development of the ambient sensor as collaborative project of The University of Alabama in Huntsville and NASA Marshal Space Flight Center, and extension of the previously published results [2]. A flexible sensor platform developed for this project features several ambient and inertial sensors, and a custom printed humidity sensor. This sensor uses a printed ceramic material as a dielectric which was developed by NASA Marshall Space Flight Center [3]. Capacitive sensors exhibit a highly sensitive response to changes in moisture with very fast response times, making it suitable for respiration monitoring.

We present respiration monitoring methods in Section 2. Section 3 presents flexible sensor platform and sensor implementation. Experimental evaluation and validation of the sensor is presented in Section 4, discussion of results is presented in Section 5, and Section 6 presents Conclusions and future work.

2. RESPIRATION MONITORING

Respiration monitoring methods can be divided into the following groups: a) monitoring of airflow, b) monitoring

978-1-7821-2734-7/20/$31.00 ©2020 IEEE

lung volume and chest motion, c) remote monitoring of breathing, and d) monitoring of modulation of other physiological signals [1].

Monitoring of airflow

Respiration creates flow of air that can be sensed unobtrusively by monitoring temperature, humidity, or carbon-dioxide (CO_2) in exhaled air, or sound during breathing [1], [4], [5]. The method of airflow sensing requires sensors to be placed directly under the nose or mouth of the patient to monitor the air being exhaled during respiration [4], [6]. 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. Caccami et al. [6] 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.

A humidity sensing approach was also taken in Kano et al. [7] 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.

Jovanov et al. used a couple of thermistors in front of nostrils or video processing of thermal sensitive film to monitor symmetry of breathing (left or right nostril dominant breathing). Possible applications include characterization of circadian rhythms, stress monitoring, and evaluation of effectiveness of relaxation techniques [4], [5].

Monitoring of lung volume and chest motion

Wearable sensors facilitate unobtrusive monitoring during prolonged periods. Several sensors have been developed to monitor respiration rate by detecting the movements of the body that are associated with respiration. Elastic belts placed on chest and abdomen with piezoelectric, piezoresistive, or pressure sensors can be used to monitor expansion of chest or abdomen. Pressure sensors can be also installed in bed to monitor respiration during sleep. An alternative approach is to measure the change of impedance of the thorax during expansion of the chest. Many of these methods are very precise but require controlled setup to get accurate measurements.

Merritt et al. [8] 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. [9] to characterize the apneas of patients during sleep studies. This sensor was printed onto a miniature board with Bluetooth Low Energy (BLE) communication. The microprocessor-based signal processing system was able to quantify respiration rate, apnea duration, and movement time.

Remote and ambient monitoring

Remote monitoring provides opportunity for unobtrusive monitoring of users [10]. The method is based on the statistical modeling of dynamic thermal data captured through a highly sensitive infrared imaging system.

Li et al. [11] present use of WiFi and passive radar for opportunistic monitoring of vital signs in home monitoring applications.

Monitoring of modulation of physiological signals

Respiration modulates other physiological rhythms. For example, respiration induced modulation can be observed in electrocardiogram (ECG) as change of amplitude of the R-peak and change of period between consecutive heart beats (interbeat intervals). Analysis of those changes can be used to assess respiration rate [1]. Moreover, changes in interbeat intervals, known as Heart Rate Variability (HRV), caused by breathing represent coupling between breathing and heart activity. This effect is called Respiratory Sinus Arrhythmia (RSA) [12] and can be used to assess stress of users [13].

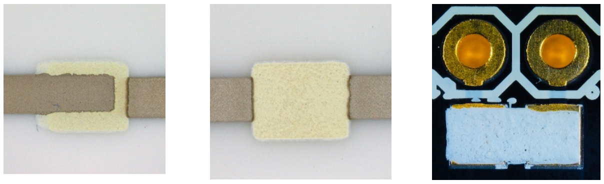
3. SENSOR IMPLEMENTATION

We implemented the custom printed sensor and used off-the-shelf temperature/humidity and CO_2 sensors to detect respiration rate.

Sensor Implementation

The sensor used in this study is made with a co-doped barium titanate ($BaTiO_3$) dielectric and gold-platinum (Au-Pt) 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. We evaluated three implementations of the custom humidity sensors using moisture sensitive dielectric [3] shown in Fig. 1.

Screen-Printed Sensor— Sintered sensors (Figs. 1a and 1b) are 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.



a) SPC sintered parallel plate capacitor
b) SCC sintered coplanar capacitor
c) PCC printed coplanar capacitor

Figure 1. Capacitive sensor implementations.

The Sintered Parallel Plate Capacitor (SPC) consists of three layers printed onto an alumina substrate in the following configuration: 1) Au-Pt base electrode 2) Co-doped BaTiO3 dielectric with a lead germinate glass binder (PbO and GeO_2) and 3) Au-Pt 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 (RH) of 40% the sensors measure between 100-200pF with response up to 1-10nF at 90% RH. Sintered Coplanar Capacitor (SCC) employs similar process, but using two parallel electrodes with dielectric between them (Fig. 1b).

Micro-Dispensed Printed Sensor Voltera— This sensor (Figure 1c) is made using the Voltera V-One printer. The printer is used to print dielectric directly on the PCB between standard PCB pads using a bulk Co-doped $BaTiO_3$ [3] dielectric printed on top. The dielectric is cured at 120ºC.

Flexible Sensor Platform

We developed a custom intelligent sensor platform that can be used to implement and evaluate a custom printed sensor. Flexible sensor platform CyBLE is shown in Fig. 2. We used prototyping area of the flexible sensor and tested implementation of two printed sensors: *thermistor* and *custom capacitive environmental sensors*.

Intelligent sensor platform is implemented on flexible Kapton substrate that allows bending of the sensor around objects (e.g. pipes and round surfaces). The platform features:

- *embedded microcontroller* Cypress CY8C4248 LQI-BL583 Arm Cortex M0 processor with 128KB of flesh memory, 16KB of RAM, BLE wireless controller, maximum frequency of 48 MHz, support for very low power modes of operation, CAPSENSE capacitive sensing (Capacitive Sigma-Delta – CSD), one channel of 12-bit AD converter, power consumption 13.4 mA @ 48MHz, and 1.5 µA in deep sleep mode,
- *On board PCB antenna,*
- *9 DOF inertial sensor* Bosch BNO080 (3-axis accelerometer, 3-axis gyroscope, 3-axis magnetometer; 6-axis sensor fusion; 9-axis absolute orientation),
- *Environmental sensor* Bosch BME280 (temperature, humidity, and pressure),
- *Low power digital gas sensor* AMS CCS811 (volatile organic compounds – VOC, and carbon dioxide CO_2),
- *power management* circuit for charging of Li-Ion batteries and regulation of voltages necessary for sensors on board, and
- *prototyping area for custom printed sensors.*

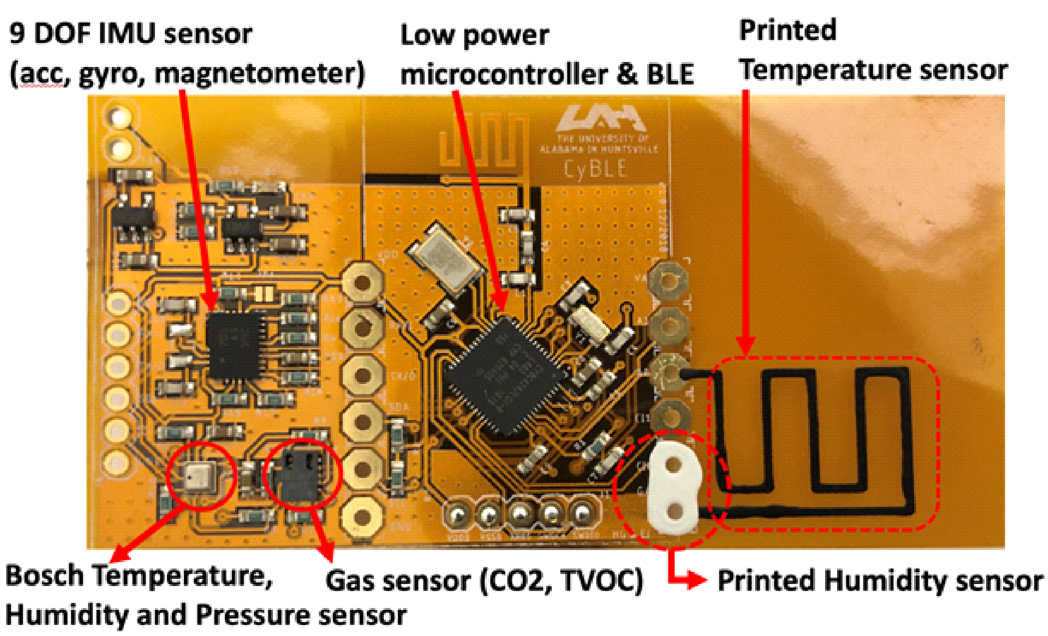


Figure 2. Flexible intelligent sensor platform CyBLE.

Flexible board features prototyping area for custom printed sensors. We tested *thermistor sensor* and *custom capacitive sensor* described in this paper.

Thermistor Sensor is implemented using carbon-based PTC resistor paste DuPont7292. Printed sensor can be used as a temperature sensor or as a heating element. We proved possible use of the sensor in both applications.

Custom Capacitive Sensor is printed using custom developed NASA dielectric ink. The sensor changes capacitance as a function of temperature and humidity of the environment.

Capacitive sensing

Capacitive sensing is very important for implementation of touch sensing and human-computer interfaces in embedded systems. Therefore, current generation of microcontrollers support direct measurement of capacitance on several pins. Microcontrollers, such as NXP MK20DX256VLH7 Cortex-M4 [14], support capacitive measurements using Touch Sense Input (TSI) interface supported on 12 pins that allows up to 12 capacitive sensors connected directly to the microcontroller without signal conditioning hardware. Sensor accuracy is in the order of tens of femto Farads [fF], depending on the tradeoff between accuracy and range. We demonstrated that quality of capacitance-to-digital converters on standard pins is sufficient even for monitoring of capacitive sensors and detection of vital signs [15].

We present fundamentals of capacitive measurement on microcontrollers we used in our experiments as they are essential to understanding operation of the sensors. Traditional impedance measurement using LCR meters allows measurement of the complex impedance at precisely defined frequency. However, effective frequency of the TSI interface depends on the measured capacitance. In addition, capacitance of our sensors also depends on frequency and that must be taken into account if absolute value of measured capacitance is important for the given application.

Detailed description of the TSI interface and configuration can be found in Chapter 50 of Reference Manual [16]. Principal organization of the TSI interface and measurement controller is shown in Fig. 3.

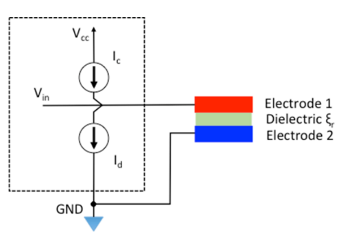


Figure 3. TSI interface and measurement setup.

One terminal of the measured capacitive sensor is connected to the ground and the other terminal is connected to one of pins with TSI interface. Internal controller uses charge current source (I_c in Fig. 3) and discharge current source (I_d) to charge and discharge external capacitor in each cycle T_c. Voltage on external capacitor during charging will be

$$V_{in} = \frac{1}{C}\int_0^t i(t)dt = \frac{1}{C}\int_0^t I_c dt = \frac{I_c t}{C} \quad (1)$$

Voltage on external pin during a single charge/discharge cycle will change as presented in Fig 4. V_{t1} and V_{t2} represent internal thresholds that determine start and end of charge/discharge periods, and ultimately length of the cycle time T_c.

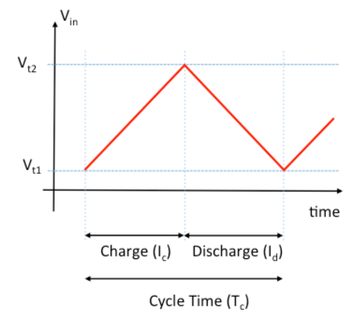


Figure 4. Change of input voltage during capacitance measurement with cycle time Tc.

From (1)

$$\Delta V = V_{t2} - V_{t1} = \frac{I_c}{C}\frac{T_c}{2} \quad (2)$$

for $I_c = I_d$ cycle time is equal to

$$T_c = \frac{2C \cdot \Delta V}{I_c} = k \cdot C \quad (3)$$

and fundamental frequency of the waveform is equal to

$$F_c = {}^{1}\!/_{T_c} \quad (4)$$

Therefore, **frequency of the signal used for measurement is proportional to the measured capacitance!** If the measured capacitance is a function of the measurement frequency *f*, we can adjust measurement frequency by changing I_c and ΔV according to (2).

TSI uses internal oscillator as a reference to measure cycle time T_c, and returns count, or number of cycles of internal oscillator proportional to measured capacitance as shown in (3). Number of scans (*NSCN*) and prescale factor (*PS*) are programmable [16]. Default TSI module sensitivity is 20 fF/count.

TSI module can be configured according to application requirements. Minimum charging current I_c is 2 µA. Longer measurements (larger value of NSCN) produce smaller noise but increases measurement time that limits applicability to real time monitoring in some applications. Although sampling frequency of the breathing signal is typically in the order of 4-10 Hz, we sampled our signal at 200 Hz to evaluate spectral characteristics of the signal. Sampling at rates over 1 KHz is possible for capacitive sensors in our application.

4. Materials and Methods

Sensor Validation and Characterization

We evaluated operation of custom sensors using standard vital sign sensor Zephyr Bioharness [17]. The sensor features breathing sensor, ECG, temperature, and 3-axis accelerometer in a chest strap.

Respiration signals recorded for one minute from Zephyr chest belt, capacitance of sintered parallel plate capacitor SPC (Fig. 1a), CO_2 count from AMS CCS811, and relative humidity from Bosch BME280 are shown in Fig. 5. Sensors were placed one inch from nostrils. All signals on the plot are presented without filtering. Respiration rate during that period was 11 BPM. Sampling frequencies were 18 Hz (respiration), 4 Hz (CO_2 and humidity) and 200 Hz (capacitance).

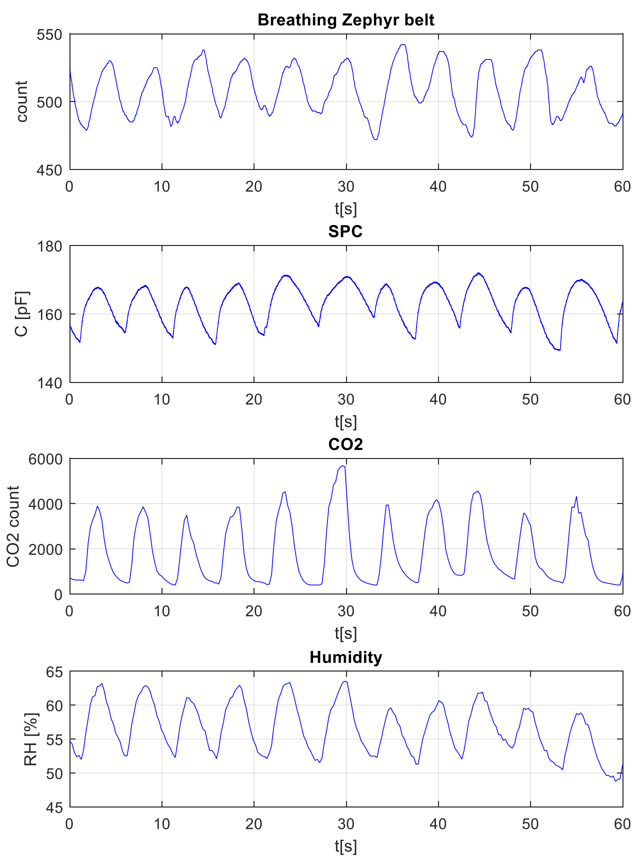


Figure 5. Breathing signals from a) chest belt, b) SPC (Fig. 1a), c) CCS811 CO_2 sensor, and d) BME280 humidity sensor.

We simulated impulse response of sensors by using intense blow over sensor to determine dynamics of response and recovery of each sensor. The results are presented in Fig. 6. Interestingly, CO_2 sensor has superior response and recovery, while humidity and capacitive sensors are saturated.

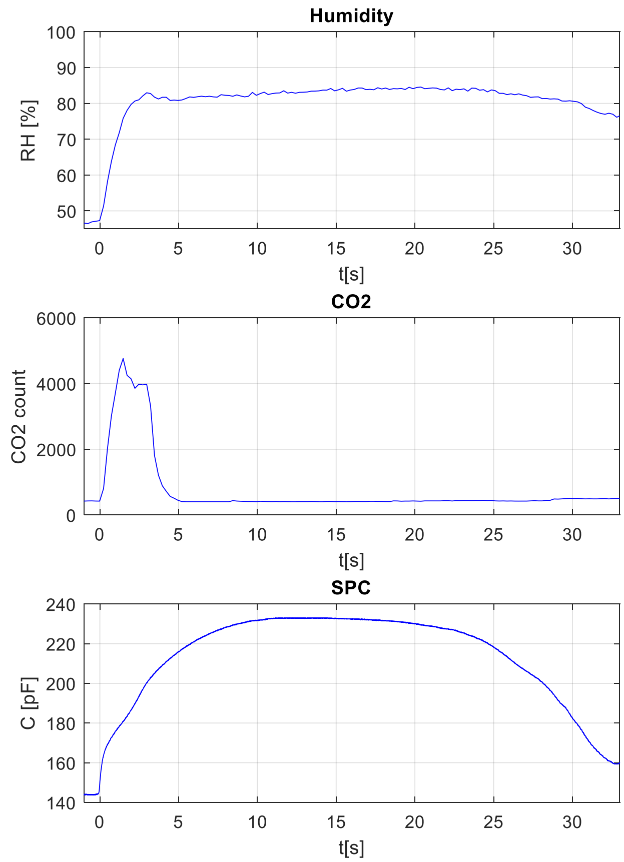


Figure 6. Simulated impulse response of sensors to intense blow directly to sensors: a) BME280 humidity sensor, b) CCS811 CO_2 sensor, and c) SPC sensor.

Experimental Validation of Sensors

We organized experiment with 9 subjects, 8 male and one female. Mean age of subjects was 29.6 years with standard deviation of 13 years. Each subject would breathe normally for one minute at the distance of 25cm followed by one minute breathing at the distance of 7.5cm.

Respiration signals were recorded from two SPC sensors placed at the distance of 7.5cm with sampling frequency of 200 Hz, CO_2 from AMS CCS811 and relative humidity from Bosch BME280 at 4 Hz, and PCS sensor at 200 Hz. Sensors were placed together with CO_2 and PCS sensors between SPC sensors. All signals were recorded using Teensy 3.2. boards [18] and sent through USB to PC for processing.

5. RESULTS

Design of the sensor represents a tradeoff between sensor sensitivity and maximum range of the capacitance to digital controller on selected microcontroller. In our previous paper, we designed the sensor to achieve maximum sensitivity at larger distances. However, total capacitance of that sensor was in the range of 1-10 nF at 90% humidity [2], while maximum capacitance that can be measured on NXP microcontroller was 200pF. Therefore, for this experiment our goal was to design sensors with base capacitance in the range of 100 pF with maximum capacitance of 200 pF. Capacitance of two SPC sensors 132.7 pF and 132 pF in room temperature conditions – RH 25% and temperature 28 °C. Capacitance of PCC was 14.9 pF (see Fig. 1). Performance of SCC sensor was significantly worse than SPC sensor and manufacturing complexity is similar; therefore, we focused our attention on comparison of SPC and PCC sensors (Figs. 1a and 1c).

Typical response of CO_2 and SPC sensors is presented in Fig. 7. This figure represents more realistic response of sensors to single breath, unlike Fig. 6 that represents forceful exhalation on sensor that is not characteristic of normal breathing. Forceful exhalation is also likely to create condensation that slows the response of the sensor (Fig. 6). It can be seen that for the normal change of RH, SPC has faster response than the CO_2 sensor, that is important for fast breathing.

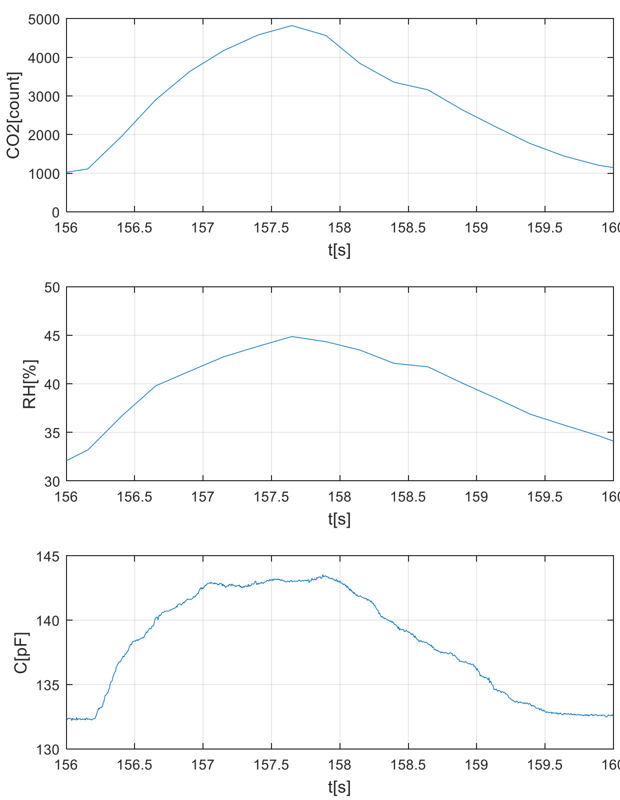


Figure 7. Change of CO_2, relative humidity, and average capacitance of two electrodes of SPC during one breathing cycle; BI=5.6s; RR=10.7BPM.

We measured the response of sensors at distance of 25 cm and 7.5 cm. New sensors SPC sensors have significantly lower capacitance than the previous generation, and their operation at distances larger than 20 cm was very unreliable. Therefore, we present only the results of processing of signals recorded at the distance of 7.5 cm.

Summary of results for SPC and CO_2 sensors for all subjects is presented in Table 1. Average breathing interval for all subjects was 4.14±0.99 s that corresponds to respiration rate of 10.7 BPM, typical for normal breathing. Average change of capacitance of the SPC sensor was 6.17±3.51 pF that is equivalent to 308 counts (or 0.3%) at default setting of the capacitive to digital controller. Relative change of the capacitance compared with capacitance in ambient temperature and humidity is 4.7%

Table 1. Summary of results for the SPC and CO_2 sensors

ID	BI[s]	std	SPC[pF]	std	CO_2[cnt]	std
1	3.86	0.32	4.12	0.66	1612.2	342.88
2	3.56	0.75	3.88	0.68	3994.2	893.05
3	5.19	0.54	4.32	1.1	5831.74	890.88
4	3.72	0.41	2.48	0.22	2992.58	366.55
5	2.6	0.24	12.12	0.92	2829.14	763.17
6	3.59	0.31	5.62	0.6	1888.52	298.32
7	3.88	0.57	4.64	1.25	2931.41	644.78
8	5.52	1.23	6.33	1.14	3563.09	422.1
9	5.35	0.5	12.02	2.3	3577.31	430.28
	4.14	0.99	6.17	3.51	3246.69	1240.22

Typical change of capacitance of the PCC is shown in Fig. 8.

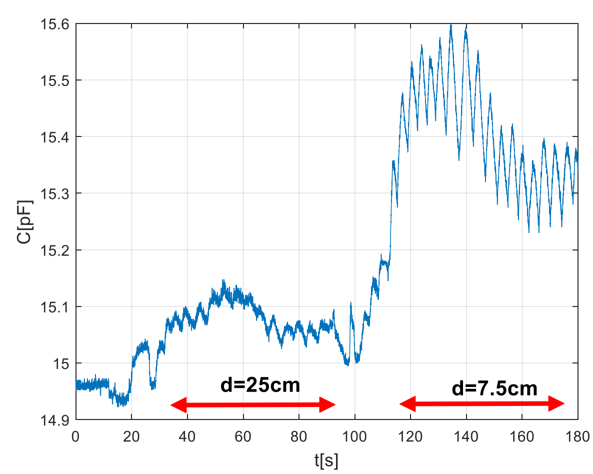


Figure 8. Change of capacitance of the PCC during breathing at distance of 25 cm and 7.5 cm; average BI=3.9 s.

Printed capacitor (PCC) had average change of capacitance of 0.55 pF, or 3.7%. Change of the average capacitance (baseline) can be clearly seen at both distances, although it is much higher at shorter distances between nostrils and the sensor.

Slower breathing rates allow sensors to return closer to the baseline level before the beginning of the new breathing cycle. Consequently, new breathing cycle will create larger change of the capacitance. This can be seen in CO_2 sensor as well. As illustration, Figs. 9 and 10 present change of the amplitude as a function of the duration of the previous breath (BI) for CO_2 and SPC capacitance.

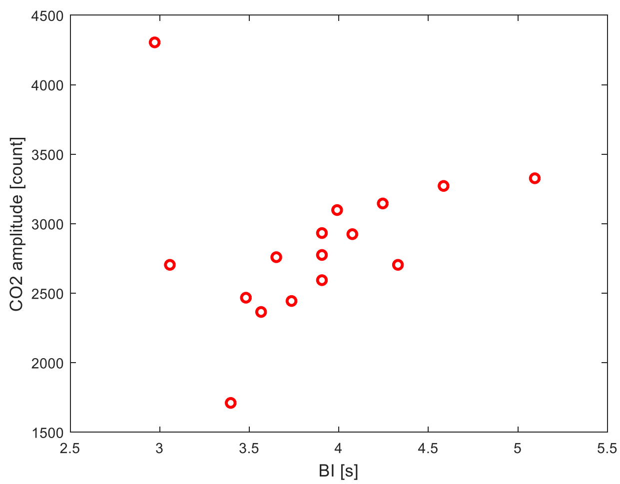


Figure 9. CO_2 amplitude as a function of the breathing interval (BI); Subject #8, Sensor distance 7.5cm.

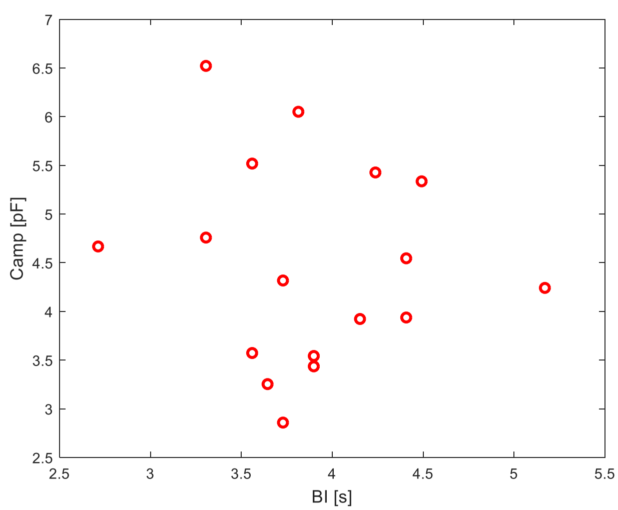


Figure 10. Change of SPC capacitance as a function of the breathing interval (BI); Subject #8, Sensor distance 7.5cm.

6. Discussion and Conclusion

Unobtrusive monitoring of human physiology and environmental conditions requires new innovative sensors and system organization. We presented the design and evaluation of the flexible sensor platform with commercially available sensors and custom developed printed sensors that could be used in space monitoring applications. Flexible sensors provide increased comfort and improved human factors. Advances in additive manufacturing, such as printed electronics, provide the following advantages:

- sensor miniaturization,
- low power or battery-less sensor implementation (e.g. RFID implementation),
- implementation on flexible, unobtrusive substrates,
- robust operation using multiple sensors,
- additive manufacturing on demand, and
- geometries or configurations can be specifically tailored to an application.

Sensor selection for a specific application must take into account several considerations concerning sensor robustness, cost, power consumption, versatility, and data quality/dynamic response. For many of these metrics the CCS811 CO_2 sensor, PCC sensor, and the SPC sensor all perform well. However, the CCS811 sensor in particular falls short in comparison to the PCC and SPC sensors in the area of power consumption and robustness – two important metrics when using a battery powered monitor around human users. The total power consumption of sampling the PCC and SPC sensors at 5 Hz on NXP processor (with an average measurement time of 1ms) is 0.5mW. Conversely, the CCS811 requires constant power and has a consumption of 90mW while being sampled at only 1Hz. Additionally, if more sensors are desired, 12 of the PCC or SPC sensors could be added to the microcontroller without an appreciable increase in power consumption; this is not the case with the CCS811 sensors which each sensor added would consume 90mW of power. The PCC and SPC sensors, being passive and based on a ceramic material, are capable of surviving and recovering from being totally submerged in water or sweat and will also be unaffected by contact with other caustic liquids. The CCS811 sensor is based on semiconductor technology and is therefore quite fragile to water, sweat, and caustic liquids and will promptly be damaged after contact with any.

Low power consumption facilitates long term monitoring with the minimum battery size, energy harvesting, or battery-less implementation. Flexible batteries can conform to application specific substrates with minimum thickness. Energy harvesting can use electromagnetic radiation, light, vibration, or temperature gradient to power the sensors. RFID technology can be used to both power and communicate with the sensors.

The PCC and SPC sensors have additional benefits in that they can be additively manufactured on demand and specifically tailored in their geometries or configurations to a specific application. Both sensor types can be implemented onto flexible substrates for use in wearable devices. Both types can also be scaled if necessary, to illicit a more desired response in particular conditions.

Unobtrusive Physiological Monitoring

Small, flexible sensors can be integrated in a spacesuit or around the habitat to facilitate unobtrusive monitoring and automatic integration of measurements. One possible application is presented in Fig. 11. Since individual sensors are directional, multiple sensors can be used to provide robust monitoring, acting as a single, distributed sensor.

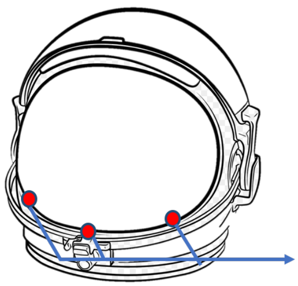


Figure 11. Unobtrusive monitoring of breathing using multiple sensors in helmet.

Since microcontrollers we tested use multiple pins to monitor capacitance (up to 12 in the case of the NXP processor), each sensor can be processed separately or all sensors can be processed as a single distributed sensor.

Unobtrusive monitoring can be also implemented using sensors in the environment and identify the user based on the vicinity of individual users. If a respiration monitor is desired to sit greater than the demonstrated 25cm, the sensors can be scaled up to have a higher baseline capacitance and a greater response in picofarads to smaller changes in humidity resulting from dispersion of the breath. Additionally, sensors can be placed in nearly any desired geometric configuration to grant symmetry or directionality to a respiration monitoring device.

Multiple sensors can facilitate monitoring of the symmetry of breath, as we reported earlier [4]. Possible applications include monitoring of circadian rhythms and stress of users that are very important for health monitoring in space.

Possible new applications may include monitoring of hydration based on humidity of exhaled breath of the users. Depending on the dynamics of implemented sensor, ambient humidity could be assessed as value between breaths, or as a value of the environmental sensors in close vicinity of the user.

The sensor could also be used in conjunction with sweat-based biometric sensors to detect the presence and possibly volume of sweat present at the acquisition point of the sensor. Such an implementation would eliminate complicated solutions to controlled fluid delivery mechanisms that often limit the functional lifetimes of sweat measuring sensors.

Change of capacitance on sensors in contact with the user may be used to assess heart rate and even heart rate variability of the users [15].

Inertial sensors implemented on our flexible sensor platform can be also used to assess activity of astronauts, with possible applications to sleep monitoring (quality and phases of sleep), behavioral monitoring, as well as frequency and pattern of use of tools and other objects.

Low Power Environmental Monitoring

Small flexible sensors can be implemented in the suite or habitat to monitor temperature, humidity, gasses, or other chemical compounds.

Possible applications include in suite monitoring, such as monitoring of relative humidity inside the suit, failure of packs, coolant loop leaks, and pump failures.

In addition to traditional environmental monitoring that includes temperature and humidity, new applications may include monitoring of leaks, urine processing leaks, and CO_2 monitoring and compensation.

Other Space Applications of Printed Sensors

Other possible applications of printed sensors may include passive sensors for monitoring of processes. For example, inductively coupled passive RLC sensor can be used to monitoring healing process of concrete curing during building of colonies and habitats. Monitoring can facilitate optimum curing and provide increased strength of the structures.

We demonstrated that thermistor ink can create not only thermistor element for measurement of temperature, but also act as a heater. Microcontrollers used in our project can generate constant current on multiple pins. Consequently, custom printed element can deliver precise amount of heat, that can be used as a part of the more sophisticated chemical sensing of specific compounds.

This paper presents implementation of a low power, flexible sensor platform and reviews possible space applications of the custom printed sensors. Our future work will include assessment of the proposed technology in specific space applications.

ACKNOWLEDGEMENTS

This project is supported in part by NASA Grant NASA/MSFC 80MSFC18N0001 *Evaluation of Intelligent Sensor Platforms for 3D Printed Electronics for Space Applications*. We would like to thank Helen Creel and Niki Werkheiser from NASA for help and support during implementation of our project.

REFERENCES

[1] H. Liu, J. Allen, D. Zheng, and F. Chen, “Recent development of respiratory rate measurement technologies,” *Physiol. Meas.*, vol. 40, no. 7, p. 07TR01, Aug. 2019, doi: 10.1088/1361-6579/ab299e.

[2] I. Small, E. Jovanov, and T. D. Rolin, “Monitoring of Respiration by Means of an Additively Manufactured Barium Titanate-based Hygroscopic Sensor,” presented at the IEEE SouthestCon 2019, Huntsville, AL, USA, 2019.

[3] T. D. Rolin and I. K. Small, “United States Patent: 9987658 - Method of manufacturing a humidity sensing material,” 9987658, 05-Jun-2018.

[4] E. Jovanov, D. Raskovic, and R. Hormigo, “Thermistor-based breathing sensor for circadian rhythm evaluation,” *Biomed. Sci. Instrum.*, vol. 37, pp. 493–497, 2001.

[5] M. L. Johnson, P. A. Price, and E. Jovanov, “A New Method for the Quantification of Breathing,” in *2007 29th Annual International Conference of the IEEE Engineering in Medicine and Biology Society*, Lyon, France, 2007, pp. 4568–4571, doi: 10.1109/IEMBS.2007.4353356.

[6] M. C. Caccami, M. Y. S. Mulla, C. Di Natale, and G. Marrocco, “Wireless monitoring of breath by means of a graphene oxide-based radiofrequency identification wearable sensor,” in *2017 11th European Conference on Antennas and Propagation (EUCAP)*, 2017, pp. 3394–3396, doi: 10.23919/EuCAP.2017.7928355.

[7] S. Kano, Y. Dobashi, and M. Fujii, “Silica Nanoparticle-Based Portable Respiration Sensor for Analysis of Respiration Rate, Pattern, and Phase During Exercise,” *IEEE Sens. Lett.*, vol. 2, no. 1, pp. 1–4, Mar. 2018, doi: 10.1109/LSENS.2017.2787099.

[8] C. R. Merritt, H. T. Nagle, and E. Grant, “Textile-Based Capacitive Sensors for Respiration Monitoring,” *IEEE Sens. J.*, vol. 9, no. 1, pp. 71–78, Jan. 2009, doi: 10.1109/JSEN.2008.2010356.

[9] S. Milici, A. Lázaro, R. Villarino, D. Girbau, and M. Magnarosa, “Wireless Wearable Magnetometer-Based Sensor for Sleep Quality Monitoring,” *IEEE Sens. J.*, vol. 18, no. 5, pp. 2145–2152, Mar. 2018, doi: 10.1109/JSEN.2018.2791400.

[10] R. Murthy and I. Pavlidis, “Noncontact measurement of breathing function,” *IEEE Eng. Med. Biol. Mag.*, vol. 25, no. 3, pp. 57–67, May 2006, doi: 10.1109/MEMB.2006.1636352.

[11] W. Li, B. Tan, and R. Piechocki, “Passive Radar for Opportunistic Monitoring in e-Health Applications,” *J. Transl. Eng. Health Med.*, vol. 6, no. 1, Jan. 2018.

[12] Zhe Chen, E. N. Brown, and R. Barbieri, “Assessment of Autonomic Control and Respiratory Sinus Arrhythmia Using Point Process Models of Human Heart Beat Dynamics,” *IEEE Trans. Biomed. Eng.*, vol. 56, no. 7, pp. 1791–1802, Jul. 2009, doi: 10.1109/TBME.2009.2016349.

[13] E. Jovanov, K. Frith, F. Anderson, M. Milosevic, and M. Shrove, “Real-time Monitoring of Occupational Stress of Nurses,” in *Proceedings of the 33rd Annual International Conference of the IEEE Engineering in Medicine and Biology Society, EMBS*, Boston, MA, USA, 2011, pp. 3640–364.

[14] “NXP Kinetis K20-72 MHz MCU ArmCortex-M4 Core.” [Online]. Available: https://www.nxp.com/products/processors-and-microcontrollers/arm-based-processors-and-mcus/kinetis-cortex-m-mcus/k-seriesperformancem4/k2x-usb/kinetis-k20-72-mhz-full-speed-usb-mixed-signal-integration-microcontrollers-mcus-based-on-arm-cortex-m4-core:K20_72?fpsp=1&tab=Documentation_Tab.

[15] E. Jovanov, “Vital Sign Monitoring Using Capacitive Sensing,” presented at the 40th Annual International Conference of the IEEE Engineering in Medicine and Biology Society, Honolulu, HI, USA, 2018, pp. 5930–5933.

[16] “K20 Sub-Family Reference Manual.” [Online]. Available: https://www.nxp.com/docs/en/reference-manual/K20P64M72SF1RM.pdf.

[17] “Zephyr BioHarness BT.” [Online]. Available: https://www.zephyranywhere.com/system/components. [Accessed: 17-Oct-2019].

[18] “Teensy 3.2, PJRC Store.” [Online]. Available: https://www.pjrc.com/store/teensy32.html.

BIOGRAPHY

Emil Jovanov *(M’98, SM’04, Fellow ’20) received the Dipl.Ing., M.Sc., and Ph.D. degrees from the University of Belgrade, Belgrade, Serbia, in 1984, 1989, and 1993, respectively. He has been with The University of Alabama in Huntsville since 1998. His research interests include wearable health monitoring, IoT (Internet of Things), wireless and sensor networks, ubiquitous and mobile computing, and biomedical signal processing. His recent work includes collaboration with NASA Marshal Space Flight Center on the development of printed sensors and intelligent sensor platforms for 3D printed electronics for Space Applications. He is recognized as the originator of the concept of wireless body area networks for health monitoring and is one of the leaders in the field of wearable health monitoring. Dr. Jovanov is a Senior Member of IEEE, and serves in IEEE EMBS Technical Committee on Wearable Biomedical Sensors and Systems. He is a member of the Conference Editorial Board and Theme 7 Editor (Biomedical Sensors and Wearable Systems) of IEEE Engineering in Medicine and Biology Society, Associate Editor of the IEEE Open Journal of Engineering in Medicine and Biology, IEEE Access, IEEE Journal of Biomedical and Health Informatics, and IEEE Transactions on Biomedical Circuits and Systems. He published more than 200 papers, 17 book chapters, and 8 U.S. patents.*

Ian Small *is an avionics failure analyst and subject matter expert at NASA Marshall Space Flight Center in the area of additive manufacturing of electronics. He graduated from the University of Colorado in 2014 with a degree in Electrical Engineering and is currently pursuing his Master's degree in Electrical Engineering at the University of Alabama in Huntsville. Ian began working at NASA in 2012 as a Co-Op and joined full time after graduation in 2014. Working with NASA's In Space Manufacturing project, Ian is currently working towards developing additively manufactured electronic devices and systems that can be produced in a microgravity or extraterrestrial environment to help NASA achieve its goals for manned deep space exploration.*

Terry D. Rolin *received his B.S. in Chemistry in 1989 and his Ph.D. in Material Science in 1993 from the University of Alabama in Huntsville. After graduating he worked with Universities Space Research Associates where he was a Project Scientist for the Crystal Growth Furnace experiment that flew aboard Columbia during the USML-2 mission. For that mission he served as a Co-Investigator on a crystal growth experiment with MIT Scientist Dr. Manfred Licthensteiger where they successfully demonstrated the demarcation of a crystal growth interface using current pulses. Following his USRA experience he worked 10 years with Nichols Research Corporation modeling heat transfer and fluid dynamics using Cray supercomputers. He is currently employed by NASA-Marshall Space Flight Center as an avionics failure analyst. His current research interests are in materials processing of nanomaterials for 3D printing avionics. He has several publications and holds 5 patents in the area of printable inks, sensors, and energy storage materials.*

Harsha Ganegoda *is a graduate student in the Department of Electrical and Computer Engineering at The University of Alabama in Huntsville. He received his Bachelor's degree in field of Computer Engineering from the University of Peradeniya, Sri Lanka in 2015. He has been working in the industry as an Embedded Research and Development Engineer over 3 years. Harsha's research interests span the areas of designing and development of real-time embedded systems for physiological monitoring, vital signal processing, wireless sensor integration in Internet of Things domain and analysis of power attacks in embedded systems. His current work includes the development of intelligent sensor platforms for Space Applications with NASA Marshal Space Flight Center.*

Mr. Curtis Hill *is a Subject Matter Expert Sr. Materials Engineer at NASA Marshall Space Flight Center in Huntsville, AL. He is leading the technical development of advanced materials and processes for In Space Manufacturing (ISM) applications. Curtis has developed a number of advanced functional materials and processes for NASA, with numerous awarded and pending patents. His research has included the development of high-performance dielectric materials for ultracapacitors and supercapacitors for energy storage and battery replacement, as well as for printed ultracapacitors. His current work for In Space Manufacturing includes the development and implementation of a new range of passive sensing technologies for wireless sensing for astronaut crew health.*