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A Flexible Capacitive Pressure Sensor for Wearable Respiration Monitoring System

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Abstract—**This paper presents the design, fabrication, and characterization of a wearable capacitive pressure sensor for respiration-monitoring systems. For the dielectric layer of the proposed capacitive sensor, Porous Ecoflex with a porosity of ~36% was prepared from a manually made sugar cube via a simple melting process. A Polydimethylsiloxane (PDMS) based Silver nanowire (AgNWs) and Carbon fibers (CFs) thin films were used for the sensor electrodes. The fabricated flexible pressure sensor exhibited a high sensitivity of 0.161 kPa^{-1} for low pressure regime (< 10 kPa), a wide working pressure range of < 200 kPa, and a high durability over 6,000 cycles. Since the proposed sensor is flexible and resizable, it can be integrated into clothes and easily placed at any location of the human body. Finally, the practicality of the sensor was successfully demonstrated by integrating the sensor into a waist belt to monitor the real-time respiration signal of the human being. The finding is highly useful to monitor respiration signal for the detection of diseases such as sleep apnea, asthma and others.**

***Index Terms*—Capacitive pressure sensor, Belt type respiration sensor, PDMS based flexible electrode, Porous Ecoflex, Real–time monitoring, Wearable pressure sensor**

I. INTRODUCTION

RESPIRATION monitoring in daily life enables health status monitoring such as the detection of diseases like sleep disorder and asthma, as well as body condition and sports applications. Sleep apnea and asthma are the most common respiratory disorders in the world [1], [2]. Sleep apnea is a sleep disorder characterized by pauses in the breathing or periods of shallow breathing during sleep. It can cause chronic fatigue, insomnia, cardiac disorder, and heart and lungs diseases. Therefore, the early diagnosis of sleep disorders is particularly important. Sleep apnea can be diagnosed by multi-parametric test known as polysomnography [3], [4]. Apnea occurs while the patients are sleeping, therefore, it is difficult for the patients to recognize themselves. In addition, when the patients stay in a hospital for an accurate diagnosis, it is inconvenient to measure the pressure, pulse, respiration, etc. overnight [5], [6]. Therefore, it is required to use a wearable device which can be easily worn at home and monitor the respiration signal during night time. Asthma is a state in which the bronchial tubes in the lungs become very sensitive and that exhibits symptoms such as increase in cough, shortness in breath, and occasionally narrowing of the bronchial channels that can cause clangor [7]. It is one of the most common chronic diseases all over the world. Around 300 million people from the world are suffering from Asthma and around 250 thousand people die every year because of it. Patients with Asthma have symptoms such as wheezing, dyspnea and coughing due to the closure of the respiratory tract, which is worst at night than during the day. Asthma is the leading cause of chronic illness in children and can also begin at any age. Most asthmatic patients are diagnosed more than two years after the onset of symptoms. Although more than half of asthmatic patients are mild, they may fail to receive appropriate treatment and may lead to fatal seizures with severe symptoms. Therefore, early diagnosis of asthma is very important.

A great amount of research has been done to develop the respiration-monitoring system using various sensors for preventing those kinds of diseases. The bed type sleeping monitoring systems have been proposed where it is assembled on the bed, but it can be applied only during the sleep time [8] - [11]. Similarly, the chest belt type system requires the wearing of an extra chest belt which is inconvenient to use in the daytime. And also, in order to diagnose the disease, breathing activity needs to be monitored for a long time. On the other hand, if such sensors are attached on the skin for a long time, the patients may feel uncomfortable and also they have to wear it attentively. To overcome these drawbacks, several noncontact type wearable sensors have been developed recently. In most of the cases, it is necessary to wear an additional chest belt, whereas, our approach does not need any extra chest belt since the sensor has been embedded in a waist belt. [12], [13]. Nowadays, flexible pressure sensors are becoming popular because they can be applied to a variety of systems including the touch on flexible displays [14], [15], soft robotics [16], [17], health monitoring [18] - [20], energy harvesting [21] - [23], wearable electronics [24], [25], and electronic skin [26] - [29]. The major advantage of a flexible capacitive pressure sensor is that it can be easily integrated into wearable devices. These flexible pressure sensors have various operating principles such as triboelectric [22], [26], piezoelectric [30], [31], optical [32],

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[33], magnetic [34], piezoresistive [35], [36], and capacitive [37] - [39] for the measurement the pressure. Among these, the capacitive pressure sensors are less sensitive to temperature, humidity, consume less power, and have a higher repeatability.

In this work, a waist belt type flexible capacitive pressure sensor was developed in facile and cost-effective approach. As fabricated sensor exhibited outstanding performance with a high sensitivity of 0.161 kPa^{-1} for low pressure (< 10 kPa), a wide working pressure range up to 200 kPa, and a high durability of over 6,000 cycles. In addition, the sensor's real-time performance was measured by integrating into a waist belt for monitoring and analysis of respiration signal.

II. Design

The sensor integrated into a waist belt along with a capacitive to voltage converter detects the real-time respiration signal [9], [10], [13]. For the measurement of respiration signal, the subject simply needs to wear the waist belt on his/her body. An illustration of the flexible capacitive pressure sensor attached on the top of waist belt is shown in Fig. 1 (a). The inset of Fig. 1(a) corresponds to the schematic of the sensor electrodes and porous dielectric layer along with its physical dimensions. The block diagram shown in Fig. 1(b) corresponds to each of the elements of signal conditioning module.

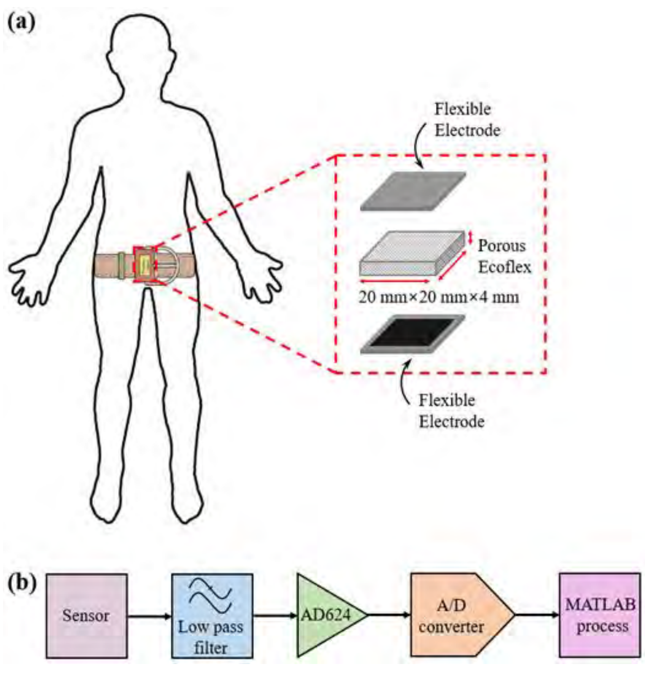

Fig. 1. (a) Diagram illustrating the sensor attached on the waist belt. The inset shows the enlarged view of the sensor with flexible electrodes and porous Ecoflex dielectric along with physical dimension and (b) signal conditioning module for output signal.

The working mechanism of the proposed sensor is shown in Fig. 2. It works on the capacitive transduction principle. The general equation governing the capacitance, C of the capacitor is given by,

$$C = \varepsilon_o \varepsilon_r A/d \qquad (1)$$

where, ε_o is the dielectric constant of the free space, ε_r is the relative dielectric constant of the material, A is the area of the electrode and d is the separation between electrodes. As the sensor is attached at the back side of the buckle, when the subject exhales, the diaphragm expands and the stomach depresses (condition of Fig. 2(a)). During this state, the sensor is not compressed, so the distance between the electrodes remains as unchanged. An effective dielectric constant, which is the linear combination of air and Ecoflex will be low since vacuum fraction of the porous Ecoflex is more dominant. As the dielectric constant of the Ecoflex is high (~2.5), the air gap inside the dielectric constant is close to 1, so ε_r is not that much high.

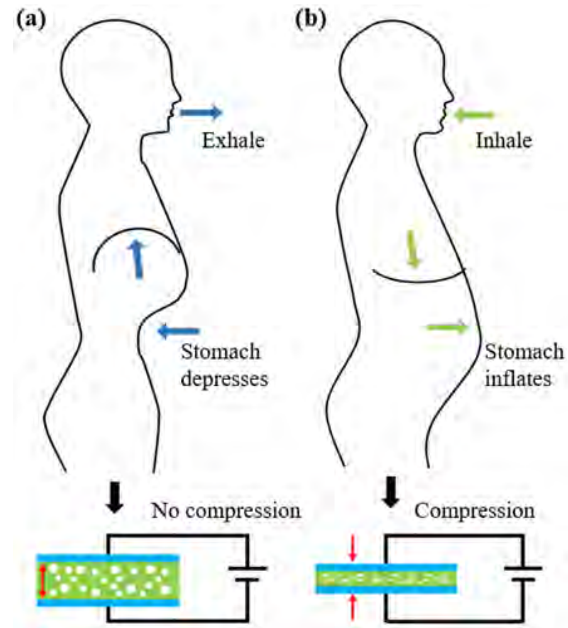

Fig. 2. Working mechanism of the proposed sensor: (a) during exhale and (b) during inhale.

When the subject inhales, the diaphragm contracts and the stomach inflates, so the sensor is compressed (condition of Fig. 2(b)). During this state, the electrodes get closer to each other thereby reducing the vacuum fraction leading to increase in ε_r. According to equation (1), whenever electrodes separation, d decreases and relative dielectric constant, ε_r increases that leads to subsequent increase in capacitance value. As the change in effective permittivity is rapid (due to two effect of parameters, d and ε_r) with respect to external pressure which leads to enhancement in the sensitivity.

III. Fabrication

A. PDMS based Flexible Electrode

PDMS based flexible conductive electrodes were prepared using AgNWs and CFs. PDMS was chosen as the substrate material because it has eco-friendliness and high stretchability among all other elastomers. AgNWs are considered as the most promising choice because Ag possesses excellent electrical

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conductivity among metals and AgNWs are known to have superior yield strength and Young's modulus over bulk Ag [40]. CFs were used to increase the conductivity of electrodes. The schematic process of fabrication of the proposed flexible composite electrode is depicted in Fig. 3(a). First, CFs with a diameter of ~7 µm were chopped into a length ranging from 1-2 mm and dispersed in Dimethylformamide (DMF) solution using an ultrasonicator. CFs dispersed suspension was poured into a Poly(methyl methacrylate) PMMA mold on the Silicon wafer and dried at 100° C for 30 min. The AgNWs solution was then poured on the top of the CF dispersed suspension and spin-coated. For the welding process, the wafer was put into 10 % Sodium chloride (NaCl) solution and then dried in oven at 95° C for 3 min until the solvent is evaporated.

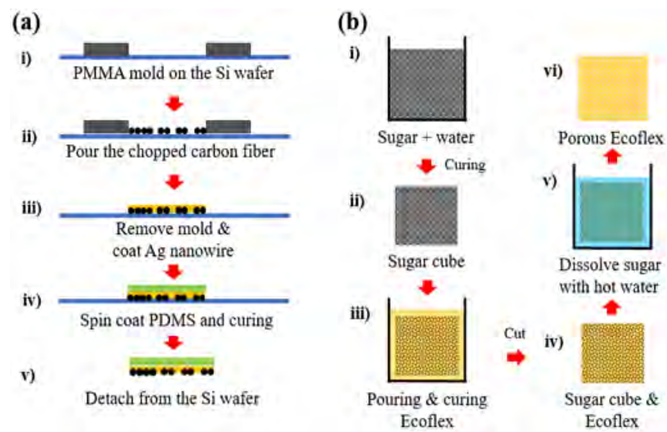

Fig. 3. Fabrication sequences of the (a) proposed flexible composite electrodes and (b) proposed porous Ecoflex dielectric.

After the evaporation of the AgNWs solvent, PDMS curing agent and prepolymer (mixed in 1:10 weight ratio) was spin-coated on the silicon wafer containing CF coating. After the curing of PDMS in the convection oven at 95° C for 3 h, the solidified PDMS film was detached from the Silicon wafer. The electrode thus obtained was cleaned sequentially by acetone, methanol, and deionized water (DI) to remove the remaining DMF solution. The fabricated electrode was cut into 20 mm × 10 mm × 1 mm (*l* × *w* × *h*). Lastly, Silver paste and epoxy were used for the wire connection.

B. *Porous Ecoflex Dielectric*

Fig. 3(b) shows the schematic drawing of the fabrication process of porous Ecoflex dielectric layer. The sugar cube was fabricated by mixing sugar powder with cold water and pouring the mixture into the PMMA mold designed into required shape. The melted sugar was put on an oven at 120° C for 1 h to obtain a hardened sugar cube. In this work, a 20 mm × 20 mm × 4 mm (*l* × *w* × *h*) sugar cube was fabricated. A base and a curing agent of Ecoflex elastomer (1:1 weight ratio) was mixed and poured on the top of the sugar cube. To cure the Ecoflex, the solution was kept under room temperature for more than 8 h. After the curing of the Ecoflex, the side walls were cut to expose the sugar, dissolved into DI water and then dried to get porous structure. The porous dielectric material was sandwiched in between flexible electrodes. The photograph image of the fabricated pressure sensor is shown in Fig. 4 (a).

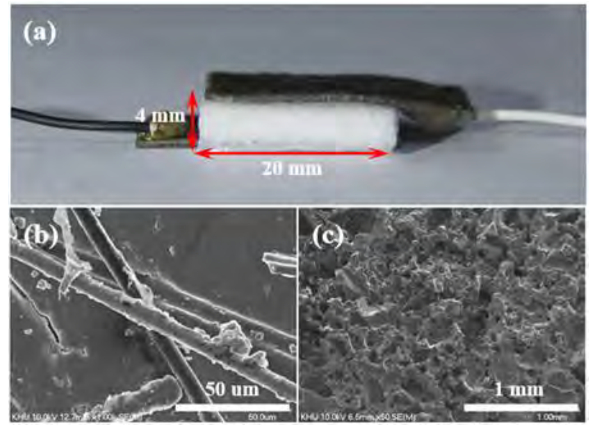

Fig. 4. Fabrication results: (a) Photograph image of the flexible capacitive pressure sensor. (b) SEM image of the flexible electrode surface and (c) SEM image of the cross-section of porous Ecoflex.

IV. RESULTS AND DISCUSSION

A. *Mechanical and Electrical Properties*

Fig. 4(b) shows the flexible electrode surface with AgNWs coated on the CFs. Using a four-point probe meter (RC2175, EDTM), the sheet resistance of the electrode was measured to be 2.5 Ω/sq. In order to enhance better conductivity and good performance of the capacitive pressure sensor, a sufficiently low value of sheet resistance is desirable. The pore characteristics of the porous Ecoflex was analyzed through scanning electron microscopy (SEM). As shown in Fig. 4(c), the pores are randomly distributed with size < 500 µm. The porosity of the dielectric was calculated to be approximately 36% from the ratio of pore volume to total volume. Pore volume was calculated by weighing the porous sample with and without water. Similarly, total volume was calculated from length, width, and height of the porous Ecoflex.

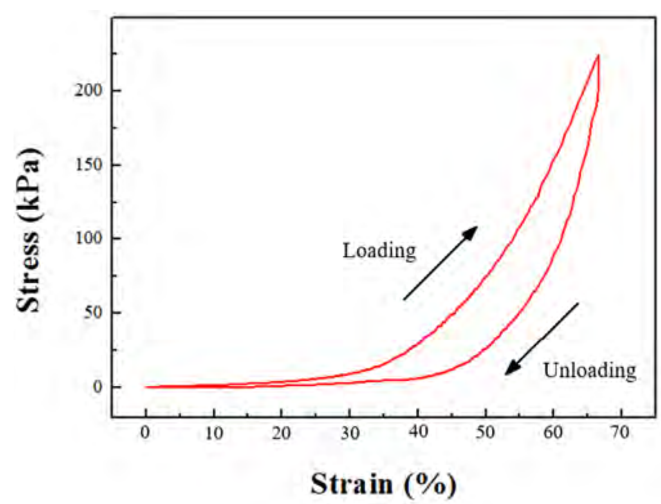

Fig. 5. The stress-strain behavior of porous Ecoflex dielectric.

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The electrical characteristics of the porous Ecoflex were measured using a combination of a moving stage (JSV-H1000, Japan Instrumentation System) and a force gauge (HF-10, Japan Instrumentation System). The compression of the sample was achieved by applying the force along z-axis. The force gauge fixture was moved at a constant speed of 1 mm/min. The force gauge attached with the moving stage measures the force associated with each compression distance. Capacitance values were measured using LCR meter (Hioki IM 3536). The sensor performance with humidity and temperature was measured by using Humidity generating chamber (ETAC, HIFLEX KEYLESS, TH403A) at 100 Hz, 1 Vpp. All the data were recorded in a computer running with corresponding software.

The stress-strain characteristic of the proposed sensor is plotted in Fig. 5. To reduce the height of porous Ecoflex from initial height of 4 mm to 1.5 mm (65 % strain), the required pressure was approximately 200 kPa. Since Ecoflex is very resilient in its nature, it is always recovered completely every time for repeated number of compression/tensile test. While unloading of the pressure back to 0 Pa, it follows almost exactly the forward path indicating low value of hysteresis error.

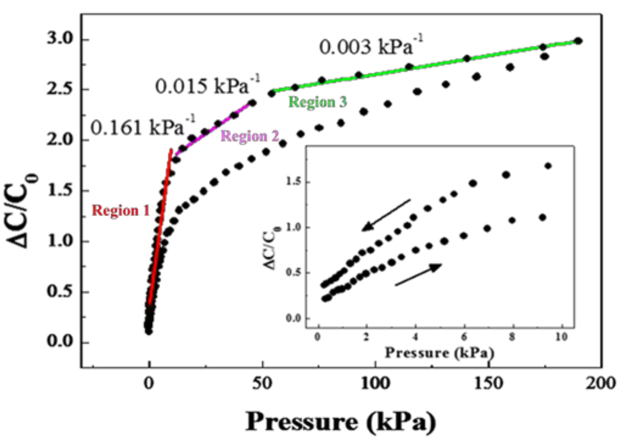

Fig. 6. Relative capacitance change under varying applied pressure.

Fig. 6 shows the relative change in capacitance for various pressure values. At the maximum compression strain of 65 %, the capacitance value was increased from an initial value of 1.24 pF to 4.94 pF. The slope of the curve in Fig. 6 defines the sensitivity, S of the capacitive pressure sensor, which is also defined by the relation,

$$S = (\Delta C / C_O)/\Delta P \qquad (2)$$

where, C_O is the capacitance when no applied pressure, ΔC is the measured capacitance change upon external applied pressure and ΔP is the change in pressure. The sensor showed three different sensitivity values up to 200 kPa. In the first region from 0 kPa to 10 kPa, the sensitivity was 0.161 kPa^{-1}. The magnified view of the sensitivity for this regions is shown in the inset of Fig. 6. From 10 kPa to 50 kPa range, a sensitivity of 0.015 kPa^{-1} was obtained, and it decreased to 0.003 kPa^{-1} beyond 50 kPa. In the high pressure region (>50 kPa), the sensor sensitivity was reduced as compared to low pressure regions since pores were already get saturated indicating no rapid change in d and ε_r.

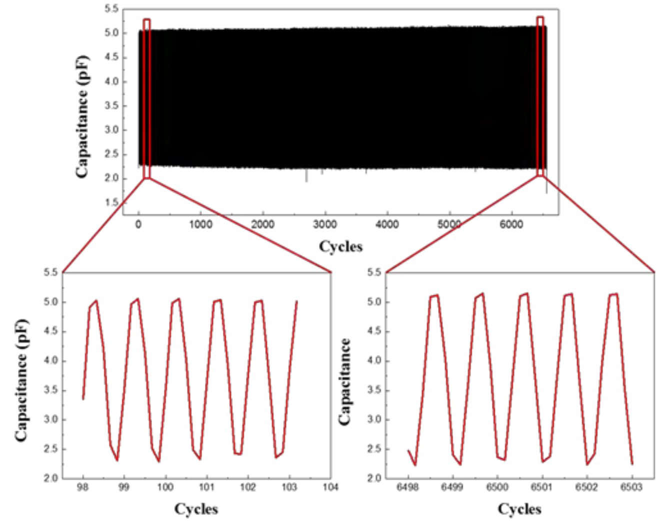

Fig. 7. Reliability test of the fabricated sensor under the repeated 6600 compression/release cycles at a frequency of 1 Hz.

The durability of the sensor was analyzed by applying repeated number of compression/release cycles. Capacitance values as a function of compression/release cycles at the frequency of 1 Hz is shown in Fig. 7. When a comparison was made between 100th with 6500th cycle as in insets of Fig. 7, there was negligible change in the capacitance values.

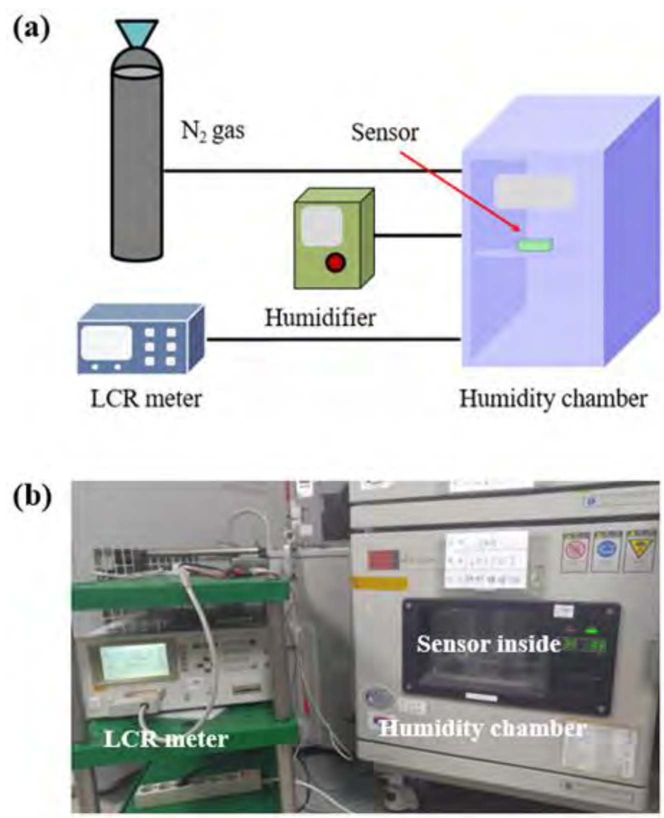

Fig. 8. Capacitance measurement under varying humidity and temperature: (a) block diagram and (b) experimental setup.

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Humidity and temperature are the main common environmental factors that can influence the performance of the sensor. To measure the dependency of the sensor with humidity and temperature, the measurements were performed by placing the sensor into the humidity generating chamber. The capacitance values of the sensor was measured using a high precision LCR meter (Agilent 4284A). The block diagram of the experimental setup and its real setup for the measurement of capacitances under various humidity and temperature is shown in Fig. 8 (a) and (b), respectively. The capacitive sensor was dried and stored in a desiccators to maintain the initial absorbed moisture level close to zero before the measurement process. The chamber humidity was increased from 25% RH to 99% RH and then decreased back to 25% RH. The sensor responded to humidity variations and the variation is directly proportional to the ambient vapor as the permittivity of the Ecoflex changes proportionally with the high dipole moments of water molecules [41]. Therefore, this humidity change was directly detected by measuring the changes in the capacitance. When water vapor is absorbed by Ecoflex, the apparent permittivity values were increased resulting in a linear increase in capacitance with Relative humidity (RH) as shown in Fig. 9. The linear slope of the trace in Fig. 9 indicates almost no significant change in sensitivity with RH. Similarly, to determine whether the temperature influenced the sensitivity of the capacitive sensor, the capacitance changes were measured at two different temperatures. Although the capacitance values were increased with temperature but the slope of the curves remained virtually unchanged, which means that the temperature does not influence the sensitivity of the capacitive sensor.

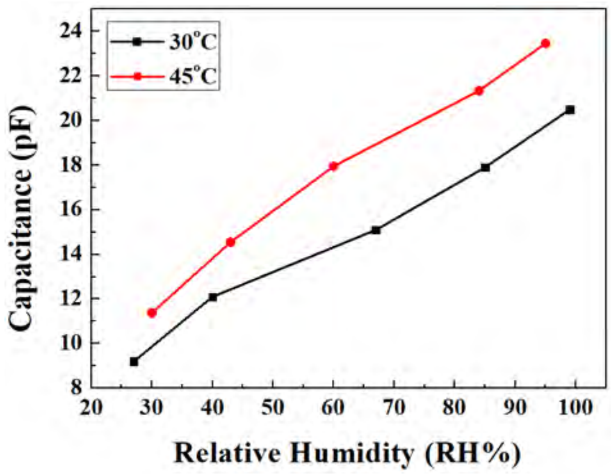

Fig. 9. Measured capacitance for different values of RH from 25% RH to 99% RH at 100 Hz with temperature 30°C and 45°C.

B. Respiration-signal monitoring system

To demonstrate the application of the flexible sensor as a smart waist belt, the fabricated pressure sensor was attached on the backside of the buckle of a commercial waist belt as shown in Fig. 10 (a). The pressure experienced by the sensor during subject's inhale and exhale were measured around 102.3 kPa (41.75 N) and 100.3 kPa (40.93 N), respectively. The pressures were measured using Load cell (model F1814 from technologies for Sensors Indicators and Systems).

Fig. 9. Measured capacitance for different values of RH from 25% RH to 99% RH at 100 Hz with temperature 30°C and 45°C.

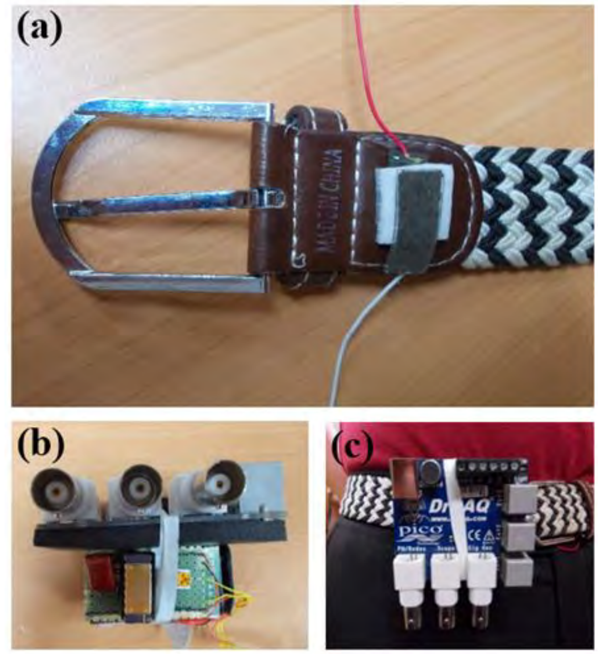

Fig. 10. (a) Sensor attached with the backside of the waist belt, (b) Low-pass filter, instrumentation amplifier and analog to digital converter and (c) whole system attached with the belt.

A flexible capacitive pressure sensor along with low-pass filter, Instrumentation amplifier (AD624), and analog to digital converter were used to transform the capacitance variations into voltage signal. A low-pass filter with a cut-off frequency of 20 Hz was used to remove high-frequency noise. DrDAQ (Pico Technology) was used as an analog-to-digital converter. Lastly, a MATLAB-coded Finite impulse response (FIR) filter was used to remove noise and interference. The Fig. 10(a) and (b) show the assembly of the components and complete system attached on a waits belt, respectively.

In this experiment, three subjects were involved during seated condition for the measurement of respiration monitoring. The subjects' information such as gender, age, height and weight is tabulated in TABLE I. First, the subject 1 in the normal state was asked to hold the inhaled breath. The measured signal for two intervals from 1−10 sec and from 25–30 sec are shown in Fig. 11(a). When the subject exercised, the more peaks occurred in the same time period. If there is respiration abnormalities, the signal shows different peaks compared with the peaks without respiration abnormalities.

TABLE I: Subjects' information

Parameters	*Subject 1*	*Subject 2*	*Subject 3*
Gender	Male	Male	Male
Age (year)	28	30	31
Height (cm)	162.6	167.7	181. 8
Weight (Kg)	62	68	82

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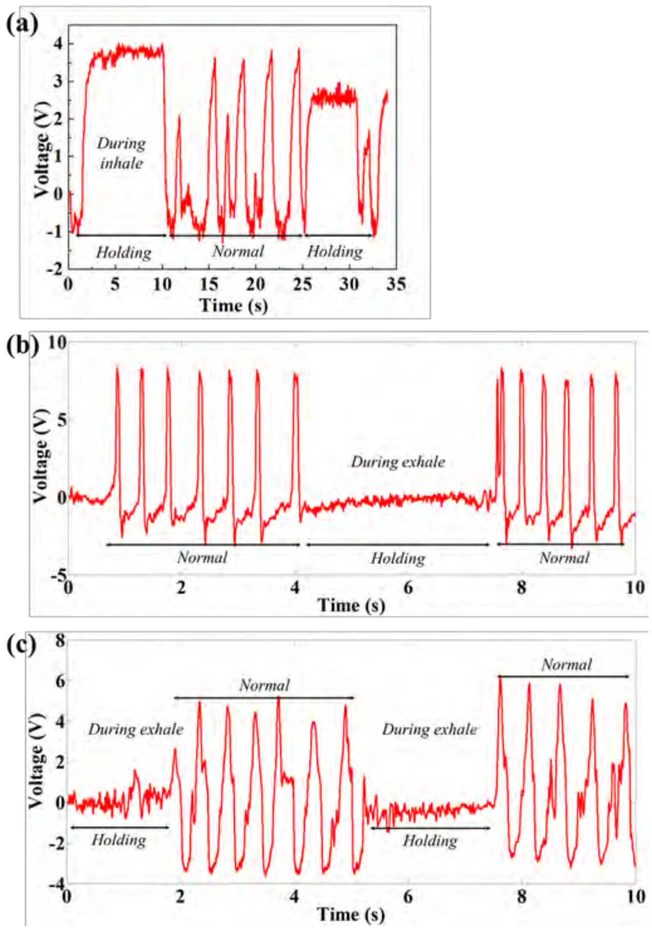

Fig. 11. Real-time measurement of respiration signal monitoring for (a) subject 1, (b) subject 2 and (c) subject 3.

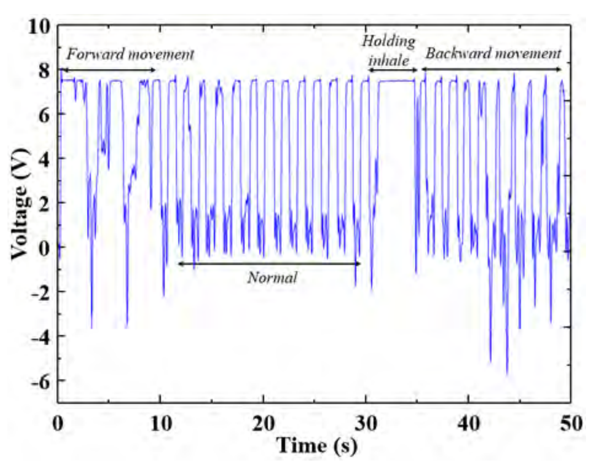

Fig. 12. Real-time performance of subject 1 bodily movement signal when bending the upper body forward and backward.

Long-time respiration-signal monitoring can therefore help in the checking of the subject's respiration state. As shown in Fig. 11(b), when the subject 2 breathed continuously but the voltage did not change during the holding of the breath for 4–7.5 sec due to exhale. Some voltage peaks were found during the normal breath from 8–10 sec. Fig. 11(c) shows the measured signal when the subject 3 holds the breath from 0–2 sec and then continues from 2–5 sec. A constantly repeated 7 peaks in 3 sec implies that the subject's stomach moves up and down 7 times. The respiration rate can thus be easily calculated from those peaks. Such signals are very helpful for the detection of the disease such as Sleep apnea and Asthma. Fig. 12 shows the converted capacitance to voltage signal when the upper body bends forward and backward during standing condition. When the upper body moves in the forward direction, few voltage peaks were found. It happened due to the movement of the sensor. But when the upper body moves in backward direction, constantly repeated voltage peaks were observed as shown in Fig. 12. It is conformed that the proposed prototype is successfully able to detect respiration signal under different situations.

V. Conclusion

In this paper, a wearable flexible capacitive pressure sensor was newly fabricated and demonstrated as a respiration signal monitoring system. The fabricated sensor was attached with the waist belt to make a non-contact real time respiration monitoring, which we successfully achieved using as fabricated sensor. Therefore, it is confirmed that the developed sensor is a solution for non-contact respiration-signal monitoring systems. It is also highly useful for a variety of wearable-electronic applications such as health-care monitoring system for an infant's breathing, Sleep apnea, Asthma, and to study the effect of sports activities.

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