Simplified Structural Textile Respiration Sensor Based on Capacitive Pressure Sensing Method

Se Dong Min, Member, IEEE, Yonghyeon Yun, Member, IEEE, and Hangsik Shin, Member, IEEE

Abstract—We propose a simplified structural textile capacitive respiration sensor (TCRS) for respiration monitoring system. The TCRS is fabricated with conductive textile and Polyester, and it has a simple layered architecture. We derive the respiration by the distance changes between two textile plates in the TCRS, which measures the force from the abdominal diameter changes with the respiratory movement. To evaluate the reliability of TCRS, both linearity test and comparison test were carried out. Three times of tensile experiment were performed to confirm the linearity of change in capacitance by the distance change. The result shows that the coefficient of determination ($R^2$) of proposed TCRS is 0.9933. For comparative study, 16 subjects were participating in the experiment. As a result, the proposed respiratory rate detection system using TCRS successfully measures respiration compared with nasal airflow detection ($R = 0.9846$, $p < 0.001$). In Bland–Altman analysis, the upper limit agreement is 0.5018 respirations per minute and lower limit of the agreement is $-0.5049$ respirations per minute. From these results, we confirmed that the TCRS could be used for monitoring of unconscious persons, avoiding the uncomfortness of subjects. Monitoring respiration using TCRS offers a promising possibility of convenient measurement of respiration rates. In particular, this technology offers a potentially inexpensive implementation that could extend applications to consumer home-healthcare and mobile-healthcare products.

Index Terms—Capacitive sensor, conductive textile, non-invasive monitoring, respiration measurement, wearable sensor.

I. INTRODUCTION

There has been increasing an importance of healthcare since our life circumstance has changed in many ways, such as the aged population society has come, the interest in an individual’s health care has increased [1], [2]. On that score, healthcare services have been developed for patient centered and high quality care technology [3], [4]. Also, there has been a move towards preventing chronic or acute diseases by early monitoring and diagnosis [2]. These changes accelerate the development of health-related products and systems that are convenient and economical. For example, research on signals from human body measurement technology without subject’s inconvenience, unconstrained and non-invasive systems that are located on the patient or user and connected to the patient or user without wire [5]–[7], becomes the key solution for the realization of ubiquitous healthcare. The requirement for such systems is that they are designed for long-term and daily health monitoring where patient or user located [7].

To guarantee user’s comfort and higher quality of life, the foregoing monitoring systems need to be based on flexible and smart technologies. Moreover, this system should enhance user’s motivation and consciousness. One of the stimulating challenges is the trial to acquire the most realistic health status in a natural environment using textile materials. This approach specifically minimizes the interaction between the sensing system and the user activity. These functions will be realized by a wearable sensor system that is integrated into clothing or installed in our living space, and this type of systems must be able to monitor a person’s vital signs and activities during daily life [8]–[10]. For this reason the use of sensing textile materials is gaining more and more attention [11].

Nowadays, several conductive fabrics have been used as a sensor and electrode in the form of fiber and yarn, and utilized to sense clothes measures the vital data with the conductive and piezoresistive characteristics of electro-conductive fabric [12]–[16]. In the most of textile sensor, textile sensing pattern are connected to signal acquisition device, and recorded signals will be stored and transmitted to other remote monitoring system. Conductive textile based measurement has already tried to measure the critical physiological signals such as electrocardiogram (ECG), electromyogram (EMG), respiration, activity pattern, and temperature [15], [16]. Moreover, the prototypes of textile-based belt-type capacitive respiration sensor were developed and have tried to be applied for respiration monitoring [13], [14].

Among these signals, respiration is one of the most critical indicators of the physiological state and it should work properly in every moment of human life because respiratory arrest could fatally damage the human body. Though this importance of the respiration, respiratory disease such as lower respiratory tract infections, chronic obstructive pulmonary disease, tuberculosis and lung cancer still remain as major ten leading causes of death worldwide. World Health Organization (WHO) reports that nearly 5 percent of the human population is suffering a respiratory disease like sleep apnea syndrome.
and about 30 percent of people in their seventies are reported to have a respiratory disease in industrialized developed lands [17], [18]. Also, according to the World Health Report 2000 of the WHO, the top five respiratory diseases account for 17.4 percent of all deaths and 13.3 percent of all Disability-Adjusted Life Years [18]. To treat respiratory disease, continuous respiration monitoring surely helpful in respiratory disease management, however reliable methods of respiration monitoring hardly exist for home and ambulatory monitoring. For this, latest approaches based on textile-based solution may be a promising approach for comfortable and practical respiration monitoring because the textile based respiration sensor is relatively low-cost, easy of use and it has simplified structure compared with current solution.

In this paper, we propose the simplified structural textile based sensor for minimal constrained respiration monitoring using conductive textile. We measure capacitance changes from the abdominal displacement in breathing and estimate inhalation and exhalation phase with signal processing. Then, estimated respiration was compared with reference signal that obtained by nasal thermocouples to validate its feasibility.

II. MATERIALS AND METHODS

A. Capacitive Respiration Sensor

The capacitive respiration sensor is implemented with a thin plate (diaphragm), usually metal or metal-coated quartz, as one plate of a capacitor [19]. It is generally built with spacing-change sensors, governed by the usual equation

\[ C = \varepsilon_0 \varepsilon_r \frac{A}{d} \]  

(1)

where \( \varepsilon_0 \) is the electric constant, \( \varepsilon_r \) is the relative static permittivity, \( A \) is cross-sectional area, \( d \) is the distance between plates. When the plates are parallel and displaced by \( \Delta d \), the change in capacitance \( \Delta C \) is

\[ \Delta C = -\frac{C}{d} \Delta d \]  

(2)

Hence as dimensions are scaled to sub-millimeter silicon size, the capacitance scales linearly, but the percent change in capacitance with displacement does not change. The capacitive respiration sensor is displacing piezoresistive pressure sensor because of lower power requirements, less temperature, and lower drift [19]. The plate is exposed to the process pressure on one side and to a reference pressure on the other. Changes in force cause it to deflect and change the capacitance. Fig. 1(c) shows the change of the distance between plates and area caused by force from inhalation. Because inhalation generates the centrifugal force, plates are pushed and get nearer. Also, centrifugal force stretch the textile sensor, therefore the area of sensor can be varied with Young’s modulus, and it makes the change of capacitance. The change may or may not be linear with force and is typically a few percent of the total capacitance. Using it to control the frequency of an oscillator or to vary the coupling of an AC signal can monitor the capacitance. It is a good practice to keep the signal-conditioning electronics close to the sensor in order to mitigate the adverse effects of stray capacitance. From the physiological characteristic, abdominal and chest muscle moves by respiration mechanism. The movement of muscle makes an amount of force to the outward. Fabricated TCRS uses this principal caused by abdominal movement, which transfers the force or pressure to TCRS. This force or pressure leads to distance change between conductive textile electrodes (see Fig. 1). Finally, inhalation and exhalation could be determined by capacitance variation derived from the changes of chest and abdominal circumference.

B. Design of Textile Capacitive Respiration Sensor

Belt type textile capacitive respiration sensor (BTCRS) consists of a basic five-layered structure forming a capacitance with a non-conducting woven interlining as shown in Fig. 2. Electro-conductive fabric, W-290-PCN, was produced by Ajin Electron (Busan, Republic of Korea). The base material of W-290-PCN is Polyester and Ni-Cu-Ni plated as shown in Fig. 3(a). Moreover, W-290-PCN is produced by an electroless plating method in order to be stronger the plating adhesive property. To achieve a characteristic of capacitor, two electric conductive fabrics have been wounded twice by a non-conducting woven interlining. Polyester woven fusible interlining (100 %) was used as an insulator between electric fabrics. Also, Fig. 3(b) shows the fabricated BTCRS, elastic fabric and snap button attached to the each end of the TCRS to fit in every size of body surface. The dimension of the BTCRS is 830 mm × 38.6 mm × 1.35 mm (width × height × thickness). The property of this material is represented in Table 1.
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Fig. 2. (a) Construction of five layered belt-type capacitive textile force sensor (BTCRS). BTCRS has repeated structure of conductive layer and insulation layer. (b) Fabricated BCTFS with snap button. Dimensing of BTCRS is 830 mm × 38.6 mm × 1.35 mm (width × height × thickness).

Fig. 3. Plating structure of electro-conductive fabric, W-290-PCN.

TABLE I
THE PROPERTY OF ELECTRO-CONDUCTIVE FABRIC (W-290-PCN)

<table>
<thead>
<tr>
<th>Parameter</th>
<th>Value</th>
</tr>
</thead>
<tbody>
<tr>
<td>Density (strand)</td>
<td>188 ± 5 Warp, 92 ± 5 Weft</td>
</tr>
<tr>
<td>Breaking strength (N)</td>
<td>671.3 ± 10 Warp, 392 ± 10 Weft</td>
</tr>
<tr>
<td>Elongation (%)</td>
<td>27.6 ± 10 Warp, 36.8 ± 10 Weft</td>
</tr>
<tr>
<td>Weight (g/m²)</td>
<td>81 ± 5</td>
</tr>
<tr>
<td>Thickness (mm)</td>
<td>0.1 ± 0.01</td>
</tr>
<tr>
<td>Surface Resistance (Ω/cm²)</td>
<td>0.005 (Min.), Less than 0.05 (Avg.), 50 (Max.)</td>
</tr>
</tbody>
</table>

Compared with previous research which uses 9 layer based on silver coated conductive textile, 3D textile and water-repellent fabric [13], proposed TCRS is assembled with lesser layers using cost-effective material.

C. Sensing Module of BTCRS

Analog front end for measurement circuit was implemented with auto-balancing bridge [20], high impedance precision rectifier [21] and 2nd order Butterworth low pass filter with 10 Hz cut-off frequency to measure the capacitance change of the textile pressure sensor. To validate the implemented acquisition module, we examine the output by changing the capacitance from 0.01 nF to 1 nF, and the result shows that the output amplitude (V) is proportionally increased with the capacitance changes. Schematic of designed circuit network is represented in Fig. 4.

D. System Configuration

BTCRS based respiration measurement system is consisted with the textile, analog front end, data acquisition module and signal processing and analysis algorithm. To record the respiratory activity, participants wear the BTCRS on their abdominal site, and the nasal thermocouple, TSD202A, Biopac System inc. (USA), is attached on the subject’s philtrum for a reference respiration measurement. The experimental setup for respiration measurement using BTCRS is shown in Fig. 5.

MP150 (Biopac, U.S.A) was used to convert the analog output of the TCRS to the digital signal and Acknowledge 3.8.1 was used for the real-time monitoring and data storing. Every data were sampled with 100 Hz sampling frequency ($f_s$). MATLAB 7.3 (The MathWorks, Inc., Natick, MA, USA) was used for a signal processing and extracting the respiratory component of the recorded signal. SPSS Statistics 20 (IBM, USA) and Sigmaplot 12 (Systat Software, Inc. USA) are used for a statistical analysis.

E. Signal Processing

Because the signal measured by BTCRS system contains frequency noise aside from the respiration signal range (0.1 Hz–0.5 Hz), we use a moving average filter to remove high frequency noise. The moving average filter operates by averaging a number of points from the input signal to produce each point in the output signal. In the equation for moving average filtering is shown in (3).

$$y[i] = \frac{1}{M} \sum_{j=0}^{M-1} x[i+j] \quad (3)$$

where $x[i]$ is the input signal, $y[i]$ is the output signal, and M is the number of points in the average. In this paper, we set the $f_s/2$ points for averaging.

Then, inflection points of inhalation to exhalation were detected by using a zero crossing method [22] with filtered signal. Inflection point detection is a key parameter in the respiration signal because peaks mean the number of breaths. Moreover, inflection point to near inflection point means a period of respiration, and respiration rate could be derived from the time interval of each inflection point.

F. Feasibility Test

To validate the feasibility of BTCRS based respiration measurement system, feasibility test was designed and performed. Sixteen subjects were participated in experiments, and there is no reported any cardiovascular and respiratory diseases. The age of the subjects ranged from 24 to 36 years old, with a mean age of 30.06 years old. The body mass index (BMI) of the subjects ranged from 19.73 to 30.09 kg/m², with a mean BMI of 24.74 kg/m². Respiration test was carried out in 3-minute duration, and the average of the respiration rates varied from 5 to 17.7 breaths per minute with a mean of 9.94 breaths per minute.

As a reference method of respiration measurement nasal thermocouple is attached near below the subject’s nose and
Fig. 4. Schematic of analog front end of the TCRS: (a) Auto-balancing bridge. (b) High impedance precision rectifier. (c) Butterworth second low-pass filter (cut-off freq. = 10 Hz).

Fig. 5. Experimental setup.

Fig. 6. TCRS characteristic curve from tensile experiment. Coefficient of determination is $R^2 = 0.9933$ for whole range. (Trend line is fitted by power law equation.)

III. EXPERIMENTAL RESULTS

A. Characteristic of TCRS

To see the capacitance change by change of force or pressure, analog push-pull gauge, FB-20K, manufactured by IMADA (Japan) was used. Three times of tensile experiment that modeled as respiration mechanism performed by increasing the force from 0 to 3 N with 0.2 N increment. This experiment is a modeling to confirm the increment of force makes two plates of capacitor closer which causes the change of capacitance. The result shows that the applied force is closely related to the capacitance change of the TCRS. Coefficient of determination by power law fitting is $R^2 = 0.9933$. These results are described in Fig. 6.

B. Characteristic of BTCRS

Respiration measurement with BTCRS is performed by raw signal measurement of abdominal movement, signal enhancement and noise removal and inflection point detection in order. Measured respiration is compared with the reference respiration method, the nasal airflow detection. Fig. 7 represents the procedure of respiration measurement using BTCRS compared with nasal thermo-
Fig. 7. Comparison of measured respiratory signals: (a) measured respiration from nasal thermocouple, (b) measured respiration from BTCRS, (c) moving average filtered nasal thermocouple signal, (d) moving average filtered BTCRS signal, (e) result of inflection point detection (nasal), and (f) result of inflection point detection (BTCRS).

Fig. 8. Comparison respiration rates between BTCRS and Nasal thermocouple (BPM: Breath Per Minute).

couple. Fig. 7(a) and (d) shows recorded raw signal from BTCRS and nasal thermocouple, respectively. In this figure, much noise is found in BTCRS compared with the raw signal measured by nasal thermocouple. Fig. 7(b) and (e) is the result of the moving average filtering of nasal thermocouple and BTCRS, respectively. Especially, in Fig. 7(e), we can find that the noise component definitely decreases in BTCRS signal. Fig. 7(c) and (f) shows the peak detection result of both signals. From these figures, we can know that respiration component can be successfully derived from BTCRS after signal processing stages.

C. Performance Evaluation

Respiration measured by BTCRS was evaluated by comparing with Nasal thermocouples in two different viewpoints; comparison of respiration rates and Bland-Altman Analysis. Respiration rate is obtained with the time differences between adjacent peaks and respiration number is obtained by counting the number of peaks. Fig. 8 shows the similarity between two different methods graphically. In this figure, we could find that the proposed BTCRS method has linear characteristic compared with the respiration of nasal airflow. As a result of paired t-test, it was shown that the correlation coefficient of measured respirations is slightly different according to subject, however the correlation coefficient was very high and significant ($R = 0.9846$, $p < 0.001$) for whole subjects.

Bland-Altman analysis is typically used to compare measurement techniques against a reference value. In Bland-Altman analysis, bias means an average difference, and limits of Agreement means 2 standard deviation that describes the range of 95% of comparison points. In our experimental result, bias is $-0.0015$ bpm, SD is $0.2568$ bpm, and limit of the agreement is $-0.5049$ to $0.5018$ bpm. Fig. 9 is the graphical representation of Bland-Altman analysis, Bland-Altman plot. From this figure, we can find that the result of the proposed BTCRS method is very similar to the reference respiration measurement technique, nasal airflow detection.

IV. DISCUSSION

From the several experiments, we try to confirm the characteristic of BTCRS and its availability for practical respiration measurement. In the first experiment, we try to find the physical characteristic of BTCRS using push-pull gauge and we found that the capacitance generated from BTCRS shows almost linear against applying pressure. From this result we confirm that the proposed BTCRS has a proper characteristic in pressure measurement in the specified range.

We carried out another experiment to make sure the availability of BTCRS in practical respiration measurement. This experiment was performed with sixteen subjects, and the result of BTCRS was compared with the result of the reference respiration method, nasal airflow detection method. To guarantee the statistical significance, measured respiration was validated its significance with paired t-test and Bland-Altman analysis. From the result of significance test, we found that the BTCRS measured respiration is highly correlated with reference method ($R = 0.9846$, $p < 0.001$). Moreover in Bland-Altman analysis we also found that the respiration rate measurements from the BTCRS are between $-0.5049$ bpm
above and 0.5018 bpm below the rate measured with the nasal thermal sensor with 95% confidence. The difference between respiration rates calculated with BTCRS signal and the nasal thermal sensor has a mean of $-0.0015$ bpm and a standard deviation of 0.2568 bpm. These results mean that the proposed BTCRS has only 0.0015 differences of breath per minute compared with reference respiration measurement technique, and the difference is allowed within around 1 breath per minute. These analyses signify that the proposed method has high similarity with reference method, and it implies that the BTCRS would be enough to apply to practical application.

### V. CONCLUSION

One key parameter in the clinical monitoring of patients is the number of breaths per minute, known as the respiratory rate. Many heart and lung diseases, especially pneumonia, affect respiratory rate. Therefore, monitoring of the respiratory rate is an important diagnostic method in planning of medical care.

This paper presents respiration measurement and monitoring with a developed BTCRS. The developed sensors are linear with sufficient resolution to measure a wide range of breathing and from different subjects. From the experimental results, proposed method shows very high significance compared with existing methods, and it implies that the BTCRS has the possibility for respiration measurement as a kind of surrogate methods. In case of belt type textile capacitive respiration sensor for breathing monitoring, it was easily allowed to every subject since it has been manufactured with a textile as cloths. Moreover, there may be performance difference in respiration monitoring by measuring position, respiration could be successfully measured in arbitrary location of abdomen because BTCRS is an elastic band, however it should be fasten.

Another advantages of the developed TCRS are cost-effectiveness. Indeed, the price of the electro-conductive fabric that used in this research, W-290-PCN, could be purchased fewer than 5 dollars per unit area (1 m²). Moreover, this design results in low manufacturing costs since all used components are available off the shelf and no special materials have to be invented unlike resistive textile sensors where specific, resistance changing materials have to be used. Therefore, it could be any pattern or shape for user friendly and be embedded into underwear, belt loops, trousers and sportswear. Furthermore, the design can be scaled to almost any size depending upon the subject’s size and where it could be located.

Most of the future work on the BTCRS lies with the improvement of the belt design as cloths which people wear everyday. Further work is required to solve the wireless instrumentation system on the sensor belt. It has been planned to use a modular wireless sensor node developed for wearable health monitoring applications.

### REFERENCES


Se Dong Min was born in Seoul, Korea, in 1975. He received the M.S. and Ph.D. degrees in electrical and electronic engineering from the Department of Electrical and Electronics Engineering, Yonsei University, Seoul, in 2004 and 2010, respectively. He is currently an Assistant Professor with the Department of Medical IT Engineering, Soonchunchung University, Asan, Korea. His research area includes biomedical sensor instrumentation and processing, healthcare sensor application, and mobile healthcare technologies.
Yonghyeon Yun received the B.S. degree in electronic engineering from the Seoul National University of Science and Technology, Seoul, Korea, in 1998, and the M.S. and Ph.D. degrees in biomedical engineering from Yonsei University, Seoul, in 2002 and 2011, respectively. He was a Research Scientist with the Korea Research Institute of Standards and Science, Deajeon, Korea, from 1999 to 2011. He was a Post-Doctoral Research Fellow with the Division of Cardiology, Severance Hospital, Yonsei University Health System, Seoul, from 2011 to 2013. Since 2013, he has been with the faculty of the Daelim University College, as an Assistant Professor of Convergence Biomedical Engineering. His current interests include medical ultrasound, biomedical signal processing and instrumentation, and u-healthcare applications.

Hangsik Shin is an Assistant Professor with Chonnam National University, Yeosu, Korea. He received the M.S. and Ph.D. degrees in electrical and electronic engineering from the Department of Electrical and Electronics Engineering, Yonsei University, Seoul, Korea, in 2005 and 2010, respectively. His research area includes biomedical signal processing, physiological modeling and computer simulation, u-healthcare and mobile healthcare technologies.