Wireless Respiratory Monitoring and Coughing Detection Using a Wearable Patch Sensor Network

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Wireless Respiratory Monitoring and Coughing Detection Using a Wearable Patch Sensor Network

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***Abstract*—Wireless body sensors are increasingly used by clinicians and researchers, in a wide range of applications such as sports, space engineering and medicine. Monitoring vital signs in real time can dramatically increase diagnosis accuracy and enable automatic curing procedures, e.g. detect and stop epilepsy or narcolepsy seizures. Breathing parameters are critical in oxygen therapy, hospital and ambulatory monitoring, while the assessment of cough severity is essential when dealing with several diseases, such as chronic obstructive pulmonary disease (COPD). In this paper, a real-time low-power wireless respiratory monitoring system with cough detection is proposed to measure the breathing rate and the frequency of coughing. This system uses wearable wireless multimodal patch sensors, designed using low power off the shelf components. These wearable sensors use a low-power 9-axis inertial measurement unit to measure the respiratory frequency, and a MEMs microphone to perform cough detection. The architecture of the wireless patch-sensor is presented. The acquisition unit, the wireless communication unit and the data processing algorithms are described. The proposed network performance is presented for experimental tests with a freely behaving user.**

***Index Terms*—Breathing rate, Coughing detection, Inertial measurement unit, Wireless, Real-Time, Low-Power, Wearable, Patch sensors network.**

I. INTRODUCTION

Health care expenses are continuously increasing year after year and taking a large part of a country's budget. During medical care, vital signs, such as heart and breathing rates, are key parameters that are continuously monitored. Coughing is a prominent indicator of several problems such as COPD, and it is also the main reason for why patients seek medical advice [1]. In fact, it is a pulmonary defense mechanism of the respiratory tract that allows the expulsion of undesirable and irritating substances. In [2], the authors studied the performance of several automatic coughing detection sensors and concluded that the best performances are achieved by systems that include an audio microphone which can also be used to measure the breathing activity [3]. The average typical healthy respiratory rate is around 12 to 20 breaths par minute. In other words, a normal breathing frequency range is around the 0.2 to 0.3 Hz. Several other methods have been used to precisely monitor breathing activity. In deed, while contactless methods exist, like the Doppler Radar [4], the ultra-wide band (UWB) radar [5] or the laser method [6], but they aren't suitable for dynamic environments since they usually require a static setup. The respiratory inductance plethysmography (RIP) is presented as the gold standard in breathing surveillance especially for wearable measurement systems [7]. It evaluates pulmonary ventilation by measuring the induction in straps attached around the chest and abdominal wall. Another wearable method is the capacitive sensor as described in [8]. It consists of integrating two textile-based capacitive electrodes on the sides of a shirt which can facilitate long-term monitoring. In [9], the authors propose a small pizeoelectric sensor placed close to the nose or mouth. It monitors the breathing flow by measuring the temperature and pressure variations. While these solutions can offer precise measurement results with stationary users, they fail with highly mobile users. Additionally, for respiratory and sleep monitoring of freely behaving users, more comfort and unobtrusivty are needed. Hence, in [10], a respiratory rate system for a stationary user is developed using the three axes of an accelerometer, and in [11], a dynamic respiration monitoring system is obtained from the fusion of an accelerometer with a gyroscope and the use of a Kalman filter, which yielded very compact systems.

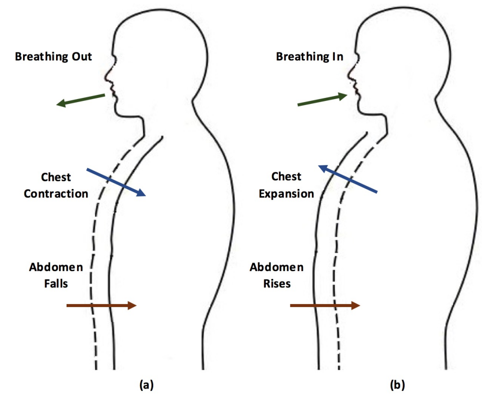

Fig. 1. Representation of the ventral body cavity, made up of the thoracic and abdominal cavities, displacement during expiration (a) and inspiration (b).

In this paper, we present a real time low-power wireless wearable measurement system based on a multimodal patch sensor network that offers unlimited flexibility and mobility for the user. It is designed with a respiratory monitoring system with a coughing detection unit. In Section II, the system design methodology is explained. In Section III and IV, the hardware

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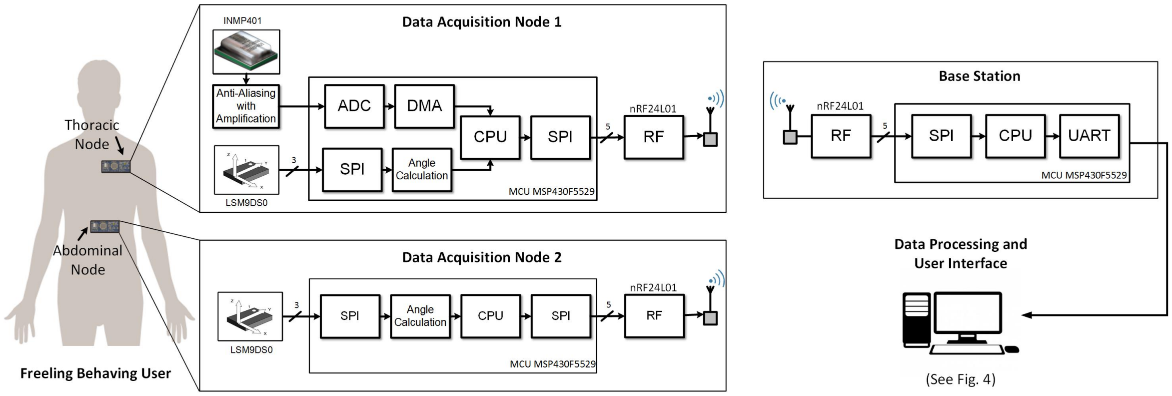

Fig. 2. Block diagram of the proposed wireless body sensor network including the data acquisition nodes, the base station, and, the data processing and user interface.

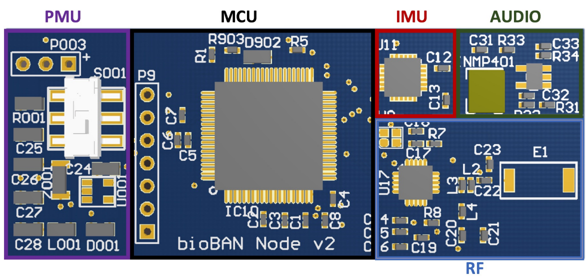

Fig. 3. Printed circuit board of the proposed multimodal patch sensor.

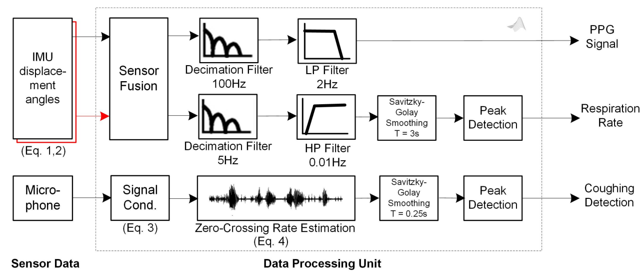

Fig. 4. Block diagram of the signal processing unit including: the respiration rate waveform unit, the respiration rate unit and the coughing detection unit.

and software system design architectures are presented. In Section V, the performance is reported for a freely behaving user before concluding in Section VI.

II. Methodology

During the respiration cycle, breathing is expressed through an upper body activity with the thoracic cage, but also with the abdominal cavity because of the important role of the diaphragm and abdominals. Especially during intense physical activities, the entire ventral cavity is compressed and expanded, as seen in Fig. 1. Thus, two inertial measurement units (IMU) can be used to measure the thoracic and abdominal cavity motions, by reading the corresponding angular motions in real-time. Data from both accelerometers and gyroscopes can be fused to obtain the breathing rate and photoplethysmographic (PPG) signals in real-time. Additionally, a small MEMS microphone is used to record the user's coughing and airway sounds, and to detect his coughing frequency.

III. System Architecture

The overall block diagram of the wireless respiratory monitoring system, presented in Fig. 2, includes two different acquisition nodes, one base station and a PC host for data processing, data management and user interaction. The data acquisition node 1, or thoracic node, is equipped with an IMU sensor and a microphone while node 2, or abdominal node, is only equipped with an IMU sensor. These acquisition nodes are responsible for acquiring data from the different sensors. They are placed such as to obtain the abdominal and thoracic breathing activities. Each one is built around a MSP430 low-power microcontroller (MCU) from Texas Instruments and use a LSM9DS0 IMU from STMicroelectronics. Node 1 is also equipped with an analog ADMP401 MEMS microphone from Analog Devices. The MCU gathers data from the IMU through an SPI interface and a 12-bit analog to digital converter (ADC) with a direct memory access (DMA), is used to sample sounds signals from the microphone after going through an anti-aliasing filter at 5 kHz. The sounds signal is sampled at a frequency of 10 kHz while the angles read from the IMU, and calculated by the MCU, are refreshed at 32 Hz, before being sent wirelessly to the base-station, using the low-power nRF24L01 radio module from Nordic Semiconductor. The latter is used for relaying all data from both nodes to the PC host where the sensor's data is processed. Finally, the acquired breathing and coughing data are extracted in real-time using the Signal Processing Toolbox in Matlab.

IV. Data Processing and Algorithms

While the abdominal and thoracic displacement angles processing are implemented directly within the sensor nodes (in-situ), the signal processing unit, depicted in Fig. 4, is implemented ex-situ, inside the PC host. In fact, through this unit is calculated the breathing frequency and the occurrence of coughing, the details of which calculations are provided in the next sections.

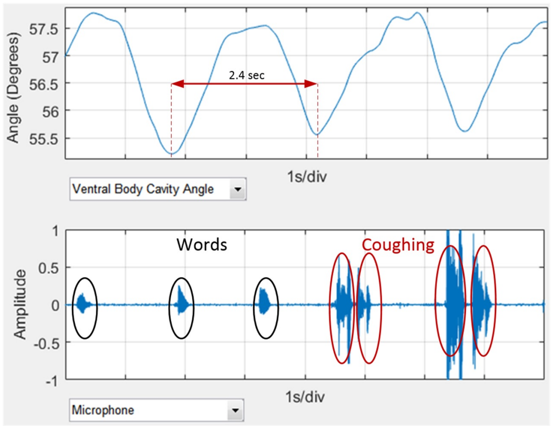

Fig. 5. Screenshot of the user interface showing the PPG (top) and the audio signal showing speech (in black circles) and coughing (in red circles).

A. User Interface

The user interface as well as the digital signal processing are both developed in Matlab. The interface allows the user to visualize the motion and the derived PPG and sound signals acquired from the 2 sensor nodes in real-time. Fig. 5 is a caption of the Matlab interface that shows the displacement of the ventral body cavity corresponding to PPG signal (Fig. 5-a), and the audio output of the microphone (Fig. 5-b). The respiratory frequency and the coughing occurrence are depicted as well.

B. Abdominal and Thoracic Displacement Angles

Within the sensor nodes, the IMU provides the accelerometer and gyroscope data to the MCU through an SPI interface link. To achieve a high speed transfer between the sensor nodes and the base station, the amount of data to be transmitted wirelessly is reduced through data processing in each node. In fact, among the axes offered by the IMU, are 3 acceleration axes and 3 rotational motion axes needed to calculate the displacement angles. Instead of sending 6 data channels to the base station, the abdominal and thoracic displacement angles are calculated thanks to a first order complementary filter at a frequency of 32 Hz within the nodes, and then sent with the rest of data. Below, (1) and (2) are used to calculate the angles when a user is standing or walking. ω_x is the rotational velocity along the x-vector, a_y and a_x are the acceleration components along the y and x-vector, α_{gyro} and α_{gyro} are the complementary filter coefficients and *angle* is the displacement angle. When the user is laying down, different vectors are used. The optimal vectors are decided by comparing them to the gravitational vector such as no disruptions occur when calculating the rotational angle during the respiration cycle since the angle is limited to -$\frac{\pi}{2}$ and $\frac{\pi}{2}$.

$$angle[n] = \alpha_{gyro}.\mu[n] + \alpha_{acc}.\nu[n] \tag{1}$$

$$with \begin{cases} \alpha_{gyro} = 1 - \alpha_{acc} \\ \mu[n] = angle[n-1] + \omega_x[n].\Delta t \\ \nu[n] = atan2(-a_y, a_z) \end{cases} \tag{2}$$

C. Breathing Activity

Breathing activity is expressed from the chest and abdomen. Hence, after initial synchronization of data coming from each node, a fusion by calculating the arithmetic mean of both, the abdominal and thoracic displacements angles, to obtain a ventral body cavity angle is performed. For the real-time PPG, only a few steps are needed. First the average displacement angle is calculated over a 3-second window and eliminated to remove the body movement, then a 20th order low-pass FIR filter with a cut-off frequency at 2 Hz is used. For the respiration rate, several steps are needed to ensure that all high-frequency components are eliminated, but also all noise artifacts coming from the body. Especially since the calculated ventral cavity angle includes the breathing movement, but also any rotation around the sensor node like body movements. Hence, the ventral cavity angle is decimated to 5 Hz to eliminate all high frequency components and followed by a 1st order high-pass filter at 0.01 Hz to remove the baseline wander. A Savitzky-Golay smoothing filter, chosen for its easy and efficient implementation, is used to smooth the signal before applying a peak detection algorithm to detect the breathing peaks. In Fig. 6 (a) and (b), the signal is presented before and after its filtering and peak detection.

D. Cough Detection

To detect coughing, we propose a simple but efficient method. Audio data is usually sampled at a high frequency of 10 kHz. In the MCU, to ensure that the data is transferred without interrupting other tasks, a DMA is used. After data synchronization, two seconds of audio data is saved for processing. The zero-crossing rate (ZCR), which is is heavily used in speech recognition, is taken as the sign change rate along the recorded audio signal [12]. Here, it is used to detect the strong important signal changes when coughing occurs. Furthermore, to maximize the algorithm efficiency, white noise is eliminated by setting to zero all data smaller than a predetermined threshold coefficient Γ, as seen in (3).

$$S[n] = \begin{cases} S[n], & \text{if } S[n] > \Gamma \\ 0, & \text{if } S[n] < \Gamma \end{cases} \tag{3}$$

The ZCR is then applied, as seen in (4), where S is the signal of length T and $1_{\Re_{<0}}$ the indicator function.

$$zcr = \frac{1}{T-1} \sum_{t=1}^{T-1} 1_{\Re_{<0}}(S_t S_{t-1} < 0) \tag{4}$$

Hence, to differentiate coughing from speech, a peak detection algorithm is applied following a Savitzky-Golay smoothing filter to detect only events with a higher zero-crossing rate.

V. Performance

This section presents the respiratory system performance. While the abdominal sensor node consumes only 12 mA, the thoracic sensor node consumption goes up to 16.2 mA with a 3.7 V supply voltage because of a higher data transmission rate due to the microphone. In Fig. 7, the consumption breakdown for the latter is shown. During the experimentation, a user had

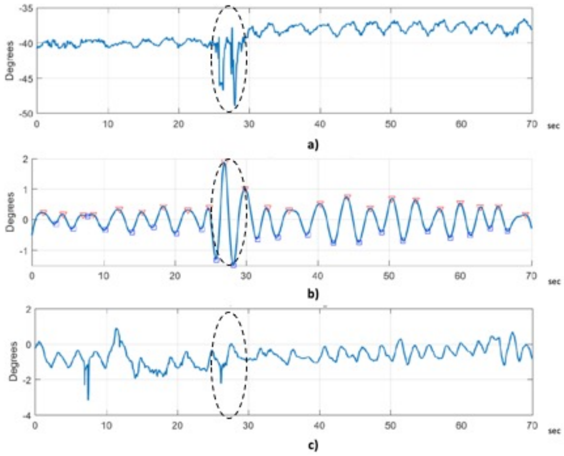

Fig. 6. The ventral body cavity angle before (a) and after (b) signal processing, and (c) the RIP signal as a reference, with a distortion circled in black

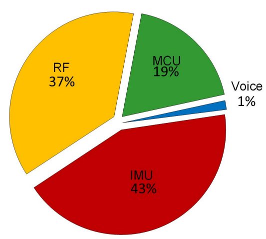

Fig. 7. Power consumption breakdown of the sensor nodes in %

the two sensor nodes placed on the thoracic and abdominal cage, and in the same time, a medical respiratory inductance plethysmography belt was attached around the chest as a reference. First, several experimental tests were done to find the optimal location, see Fig. 8. Second, a performance test was done while the user was walking to demonstrate its robustness. Finally, the proposed monitoring system's performance is compared to the RIP belt. In Fig. 6, the angle before and after filtering, and the RIP signal are all shown while the user is walking. The figure particularly shows the system functioning correctly while the user is moving. Furthermore, it is able to detect breathing patterns during heavy distortions as seen circled in black at the 25th second in the three graphs, where the user disrupts the signal by sneezing. In Fig. 5, an image of the interface functioning showing the PPG signal and the audio signal. Clearly, the cough events and the pronunciation of words can be discriminated in the audio signal. Indeed, the ZCR was able to differentiate between the different audio sounds and recognize a cough.

VI. Conclusion

A real-time wireless respiratory monitoring system with coughing detection is presented for patient surveillance during ambulatory, hospital and house care. It uses low-power electronic building blocks and is designed to maximize the movement and comfort for the user. Its set-up is much quicker and easier than the RIP used in hospitals since it doesn't need any synchronization. Finally, results show that the system can acquire breathing data while the patients is resting but also

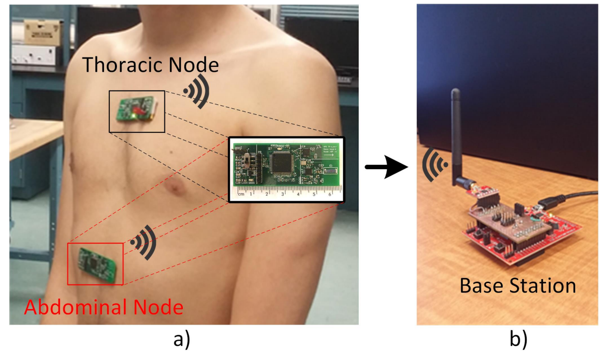

Fig. 8. Picture of the 2 sensor nodes (node 1 in black and node 2 in red) placed on user (left) and base station (right).

when walking. While the system is able to detect the coughing occurrence, more complex algorithms have been proposed that can be easily implemented to improve its reliability especially when talking [2] [12]. The MCU and RF module power consumption can be greatly improved. The system can also be expanded to include a cardiovascular monitoring unit to increase diagnostic reliability.

References

[1] Schappert SM and Burt CW. Ambulatory care visits to physician offices, hospital, outpatient departments, and emergency departements: United states 2001-02. pages 1–66, 2006.

[2] T. Drugman, J. Urbain, N. Bauwens, R. Chessini, C. Valderrama, P. Lebecque, and T. Dutoit. Objective study of sensor relevance for automatic cough detection. *IEEE Journal of Biomedical and Health Informatics*, 17(3):699–707, May 2013.

[3] Y. Nam, B. A. Reyes, and K. H. Chon. Estimation of respiratory rates using the built-in microphone of a smartphone or headset. *IEEE Journal of Biomedical and Health Informatics*, 20(6):1493–1501, Nov 2016.

[4] Xiaomeng Gao, A. Singh, E. Yavari, V. Lubecke, and O. Boric-Lubecke. Non-contact displacement estimation using doppler radar. In *Engineering in Medicine and Biology Society (EMBC), 2012 Annual International Conference of the IEEE*, pages 1602–1605, Aug 2012.

[5] Qiuchi Jian, Jian Yang, Yinan Yu, P. Bjorkholm, and T. McKelvey. Detection of breathing and heartbeat by using a simple uwb radar system. In *Antennas and Propagation (EuCAP), 2014 8th European Conference on*, pages 3078–3081, April 2014.

[6] T. Kondo, T. Uhlig, P. Pemberton, and P.D. Sly. Laser monitoring of chest wall displacement. *European Respiratory Journal*, 10(8):1865–1869, 1997.

[7] A. Hart, K. Tallevi, D. Wickland, R. E. Kearney, and J. A. Cafazzo. A contact-free respiration monitor for smart bed and ambulatory monitoring applications. In *2010 Annual International Conference of the IEEE Engineering in Medicine and Biology*, pages 927–930, Aug 2010.

[8] Subrata Kumar Kundu, Shinya Kumagai, and Minoru Sasaki. A wearable capacitive sensor for monitoring human respiratory rate. *Japanese Journal of Applied Physics*, 52(4S):04CL05, 2013.

[9] Y. Ono, D. Mohamed, M. Kobayashi, and C. K. Jen. Piezoelectric membrane sensor and technique for breathing monitoring. In *2008 IEEE Ultrasonics Symposium*, pages 795–798, Nov 2008.

[10] A. Bates, M. J. Ling, J. Mann, and D. K. Arvind. Respiratory rate and flow waveform estimation from tri-axial accelerometer data. In *2010 International Conference on Body Sensor Networks*, pages 144–150, June 2010.

[11] Ja-Woong Yoon, Yeon-Sik Noh, Yi-Suk Kwon, Won-Ki Kim, and Hyung-Ro Yoon. Improvement of dynamic respiration monitoring through sensor fusion of accelerometer and gyro-sensor. *Journal of Electrical Engineering and Technology*, vol. 1, Jan 2014.

[12] B. Ferdousi, S. M. F. Ahsanullah, K. Abdullah-Al-Mamun, and M. N. Huda. Cough detection using speech analysis. In *2015 18th International Conference on Computer and Information Technology (ICCIT)*, pages 60–64, Dec 2015.