Ambulatory Devices Measuring Cardiorespiratory Activity with Motion

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Abstract: Holter-type devices with sets of sensors, enabling long-term measurement of quantitative respiratory parameters, were designed and constructed. Pneumonitor 2 was intended for physiologic and athletic applications, and Pneumonitor 3 for sleep studies. Both allow simultaneous, comfortable, ambulatory monitoring of cardiorespiratory activity, such as *ECG*, impedance pneumography (*IP*), and motion; the second device also allows pulse oximetry and uses improved setting with combined receiving *ECG* and *IP* electrodes. Preliminary results showed that our prototypes provide signals reliable to monitor heart and breathing activity quantitatively. We tested the devices in different conditions, including walking, stair-climbing, cycle ergometer training, natural daily activity, and sleep. They can quantitatively measure respiratory flows, volumes, and minute ventilation using *IP* after calibration. They are also able to estimate tachogram from *ECG*. They allow the detection of subject activity and body position via accelerometer and gyroscope, which is helpful during *IP* calibration and interpretation. Pneumonitor 3 also enables measurement of blood saturation with a pulse wave (pulse oximetry).

1 INTRODUCTION

1.1 Traditional Respiratory Monitoring

Respiratory monitoring is mainly based on airflow measurement using spirometry and longer recording of flows and volumes with pneumotachometry (*PNT*), allow one to assess obstruction, restriction, and other issues in patients with asthma and chronic obstructive pulmonary disease (Miller et al., 2005). These methods measure flow values directly and are the most accurate. However, they are hard to use in ambulatory conditions and cannot be performed in an outpatient setting. In clinical practice, respiratory parameters are considered only for conditions which allow connection to a tube, not for natural daily and nightly functioning or athletics outside a gym.

The ability to record respiratory effort and quantitative flow- and volume-related parameters (as well as inspiratory and expiratory phases, and respiratory rate) in new settings could be of real benefit and impact both for physiologists and for sport medicine experts.

1.2 Ambulatory Respiratory Monitoring

Traditional examination captures a single point in time. From a clinical point of view, additional testing carried out under more natural conditions and taking into account activity, circadian rhythms, *etc.*, could expand early diagnosis.

Another factor is sleep; respiratory activity is known to weaken during the night (McNicholas, 1997). Basic analysis commonly performed during polysomnography consists of detection of snoring and central or obstructive sleep apnea, *e.g.*, in connection with blood saturation and heart activity (Hoyer et al., 2001; Roebuck et al., 2013). The breathing patterns are usually measured indirectly by a belt and a cannula, which make sleep less natural and comfortable.

There are some methods, which could be considered as an alternative to *PNT*, impedance pneumography, respiratory plethysmography, or acoustical approach. Based on the context of continuous studies, breathing could be precisely described by simple parameters: minute ventilation, tidal volume, and respiratory rate. All of these could be determined with

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impedance pneumography (*IP*), which seems to be the most accurate in terms of shape of the volume-related signals. It is known that this method can be used to measure ventilation in ambulatory settings (Seppa et al., 2010; Mlynczak et al., 2015). *IP* could be performed during sleep along with Holter *ECG* monitoring.

1.3 Motion and Heart Activity

It has been noted that the calibration coefficients converting impedance to volume are dependent mainly upon subjects and body positions (breathing depth and rate were of less impact) (Mlynczak et al., 2015). Therefore, the reliable calibration procedure should concern measurements performed in various body positions, and it would be worthwhile to track the current position during *IP* measurements to apply specific calibration coefficient.

Motion tracking seems also very important with regards to sleep studies, *e.g.*, for hypnogram estimation. The motion-associated artifacts in *IP* signal could be adaptively removed, smoothed, or marked using a motion signal synchronized with the respiratory one, without any cooperation from the subject.

Based on the present guidelines, each sleep study device should be equipped with pulse oximetry and heart activity registration unit (Collop et al., 2007). Furthermore, the simultaneous analysis of heart activity along with respiratory one could allow assessment of the autonomic nervous system operation (*e.g.*, autonomic heart regulation investigated from heart rate variability). Such experiments are rarely carried out under natural conditions. *Grossman et al.* (Grossman and Taylor, 2007) suggested that the depth and the frequency of breathing affect heart activity in different way and presented different physiological mechanisms of control between heart rate and heart rate variability.

1.4 Objective

The aim of the study was to prepare portable devices which would register respiratory activity (using impedance pneumography) together with *ECG*, motion, and/or pulse oximetry (saturation, pulse wave) for physiologic and sport applications, and sleep studies.

Figure 1: Pneumonitor 2 measurement device.

2 METHODS

2.1 The Devices

Pneumonitor 2 was made as a modification of the first version (Mlynczak et al., 2014), extending the device sensors (*IP*, *ECG*, and motion) and power management. It is presented in Fig. 1.

Pneumonitor 2 is $14.2cm\,x\,6.9cm\,x\,2.3cm$ and has a weight of $160g$. It is based on the tetrapolar impedance measurement method with a sinusoidal application current with an amplitude adjustable up to $1mA$, and $100kHz$ frequency.

We improved power management by replacing the elements that consumed the most energy, and by using a rechargeable battery, similar to those found in mobile phones, with $900mAh$ capacity. This allowed measurement for at least $12h$ and SD card recording, with a $250Hz$ sampling frequency. The sampling frequency is chosen as a compromise between the output data size, the possible jitter in the estimation of the R-wave fiducial point, and the accuracy sufficient from sleep studies perspective (Task Force, 1996).

ECG amplifier has the gain of $100V/V$, the bandwidth up to $100Hz$, $10nV/sqrt(Hz)$ noise and $10bit$ resolution). InvenSense's MPU-6050 (accelerometer and gyroscope unit, available commercially) was employed to estimate motion.

Wireless communication was omitted since no clinical necessity for online analysis of results was noted. A very small OLED screen was added to show signal waveforms. A simple, native PC application capable of exploratory data analysis and recording of samples in a database was also prepared.

Pneumonitor 3 had the same *IP*, *ECG*, and motion components. It differed due to the addition of a wireless pulse oximetry module (Contec CMS50EW, commercially available), connected to the main de-

Figure 2: Pneumonitor 3 device with pulse oximeter.

vice via Bluetooth communication. It had a custom-made housing, which was more solid due to the expected operating conditions (*e.g.*, regarding the possibility to connect the device to A/D converter via BNC connectors). It uses improved setting of electrodes described in the Configuration section. Pneumonitor 3 prototype is $16.7cm\,x\,10.1cm\,x\,3.5cm$ with a weight of $330g$. It is presented in Fig. 2.

2.2 Configuration

Pneumonitor 2 has 7 leads, intended to be attached using standard spot, disposable *ECG* electrodes: 4 for *IP* and 3 for single-lead *ECG*. Pneumonitor 3 has 5 leads, 2 receiving *ECG* electrode were combined with *IP* inputs - both signals could be easily separated based on different frequency spectra.

The devices provide two preset levels of internal amplification in the impedance receiving chain, in order to adjust the measuring range depending on the electrode configuration.

Pneumonitor 2 and Pneumonitor 3 are intended to be used with one of two electrode configurations. First, the one proposed by *Seppa et al.* likely provides the best linearity between impedance and volume changes; in this configuration, the receiving electrodes were placed on the midaxillary line at about 5th- and 6th-rib level and the application electrodes were mounted on the proximal side of the arm at the level of the receiving ones (Seppa et al., 2013). The second configuration is a ”classical” one, where both application and receiving electrodes are positioned at about 5th- and 6th-rib level. It is considered worse in terms of transition linearity, yet most likely optimal in terms of motion artifacts.

The voltage-to-impedance transition function was established for two settings of internal input amplification. We confirmed that the function is linearly

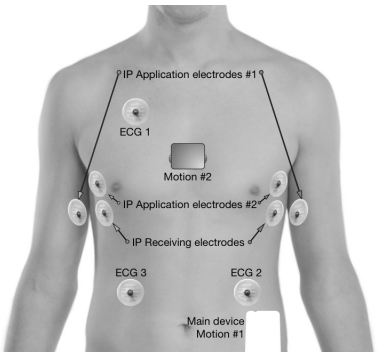

Figure 3: The scheme for positioning of electrodes for impedance pneumography and *ECG*, and of sensors for motion tracking. #1 - proposed in (Seppa et al., 2013); #2 - ”classical” one; positions of *IP* inputs are the same for both configurations; for Pneumonitor 3 they are combined with receiving *ECG* electrodes (ECG1 and ECG2).

dependent on the amplification setting.

ECG could be registered by Pneumonitor 2 using various electrode configurations. For basic measurements, we used differential electrodes placed below the right clavicle on the chest and below the left 12th rib. A reference electrode was then placed below the right 12th rib, roughly symmetric to the second electrode across the sagittal plane.

Different strategies could be applied for motion measurement using accelerometer and gyroscope. Fig. 3 shows a full schematic for placement of electrodes and proposed placement of motion sensors.

Saturation level and pulse wave were measured with the finger sensor.

2.3 Cardiac Component in Impedance Signal

We observed that electrical heart activity is a component of the raw signal registered using impedance pneumography. Despite having the option to extract the *ECG*-like signal (cardiac *IP* component), we did not remove the one-lead *ECG* unit, because:

- the ratio of amplitudes of cardiac and respiratory *IP* components in raw impedance signals changes during breathing and differs depending on the subject,
- the cardiac *IP* component is smoothed compared to the real *ECG* signal, particularly during the QRS complex,
- the cardiac *IP* component is registered with a specific electrode configuration optimized for respiratory, not cardiac recordings, and
- the one-lead *ECG* signal could be recorded for various configurations and synchronized with the

distorted *IP* signal, allowing its adaptive filtration.

2.4 Evaluation

The described devices are intended to be used in extensive study taking into account physiology-, sport- and sleep-related protocols. To validate the reliability of the respiratory- and *ECG*-related parts of the devices we performed the calibration procedure (free 30-second-lasting breathing in supine, sitting and standing), and the test procedure consisting of 6 normal breaths and then 6 deep breaths (with the subjective difference), for three breathing rates (6, 10 and 15 BPM) and for the same three body positions as during calibration (Mlynczak et al., 2014).

Based on the reference *PNT* (from Pneumotachometer M909, by Medikro Oy, Finland, and Simpson's quadrature for integration of flow values into volumes), we fitted the best linear model without intercept (after mean value removal) between *PNT* and respiratory *IP* component (after subtraction of the smoothed (respiratory-related) signal from the raw one). The calibration coefficient was used to convert the impedance signal into volumes. We evaluated the accuracy of tidal volume estimation by analysis of the difference between maximum and minimum value of volume for each breathing cycle, between reference *PNT* signal and respiratory *IP* component after applying the calibration coefficient.

We also estimated and compared heart rates from *ECG* and cardiac *IP* component signals using adaptive amplitude thresholding method (after drift removal). The *RR* intervals were calculated as a difference between two consecutive *R* points, and tachogram were transformed to beats per minute (BPM) and interpolated to $250Hz$ in order to have the same number of samples as original *IP* or *ECG*.

The preliminary pilot testing was carried out on a group of 10 participants (all males). MATLAB software was used to review and analyze the results.

In order to evaluate qualitatively the acquired signals and the usability from subjects' perspective, we also asked to carry out testing during more "natural" conditions: walking, climbing stairs, exercising on a cycle ergometer for 90 seconds with the increasing load (from $50W$ to $200W$). The activity and changing positions on a bed were tested during sleep.

3 RESULTS

Simultaneous *ECG*, *IP*, and *PNT* recordings during the static test and cycle-ergometer exercise showed that the quality of the signals allows one to calculate the tachogram, respiratory rate, and tidal volume.

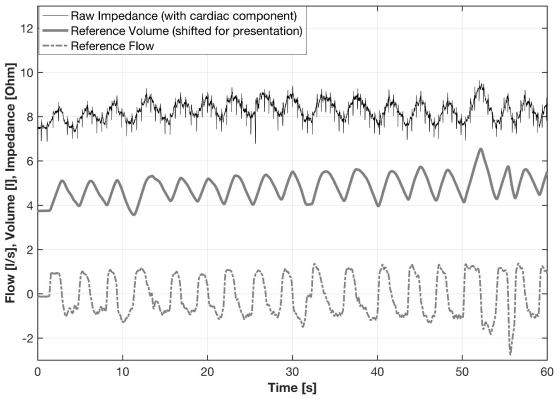

Figure 4: Sample impedance pneumography signal without removal of the *ECG* component and pneumotachometry signals, obtained during exercising on a cycle ergometer.

The mean overall accuracy of tidal volume estimation for the static test was 86.5%. Heart rate estimation based on cardiac *IP* component, for supine body position, reached mean 97.3% agreement in comparison with *ECG*.

No artifacts that might preclude analysis (without the one at the beginning of some motions) were shown during pilot evaluation in natural situations, *e.g.*, changing positions on a bed or walking.

Fig. 4 provides sample raw *IP* signal measured while exercising on a cycle ergometer (with pneumotachometry flow and volume signal). Sample recordings of *IP* and *ECG* with breathing phases and *HRV* analysis were presented in Fig. 5. Finally, Fig. 6 presents sample smoothed blood saturation data, along with pulse wave, the *IP* signal distorted with motion artifact, the 3-axis accelerometry signal and *ECG*, registered during sleep.

4 DISCUSSION

The first ambulatory system was described by *Vuorela at al.* (Vuorela et al., 2010). In contrary to their construction, we decided to remove most of the analog blocks for signal conditioning and processing and expand the analysis performed simultaneously after signal acquisition. The key novelties are adding the ability to measure blood saturation and pulse wave, and reducing the number of electrodes in Pneumonitor 3, from 7 to 5, which could be particularly useful for subjects' comforts during sleep studies.

Our prototypes will allow further research to confirm whether the use of *IP* as a clinically relevant method is possible in ambulatory conditions (in par-

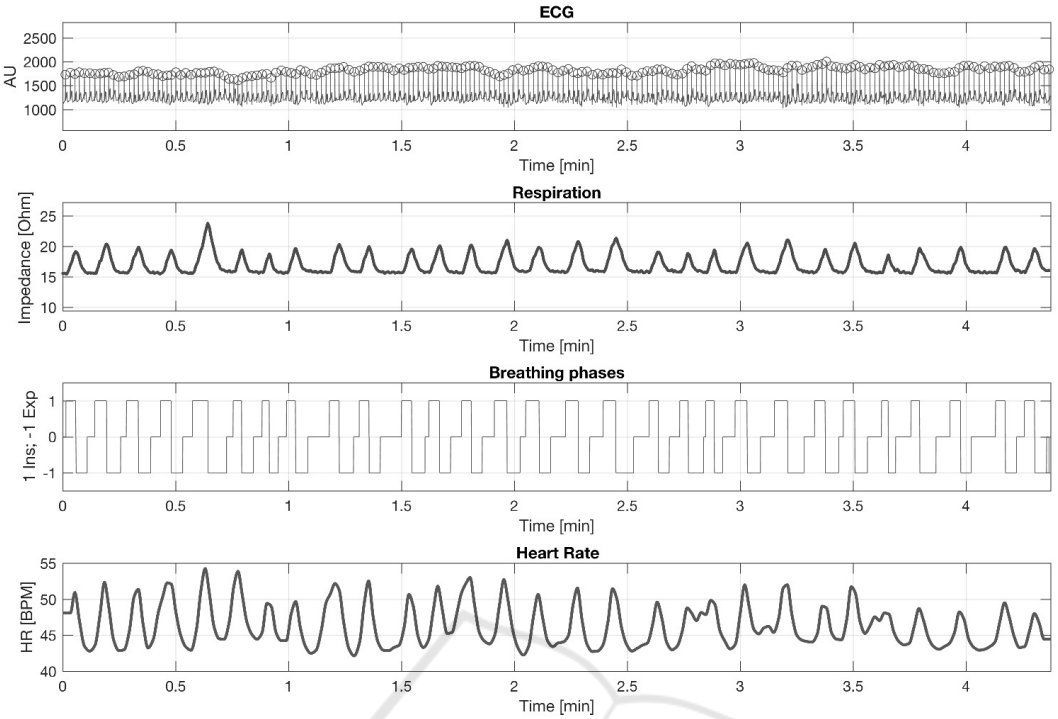

Figure 5: Sample *ECG* and *IP* signals and correspondingly calculated respiration phases and tachogram; recorded using Pneumonitor 2.

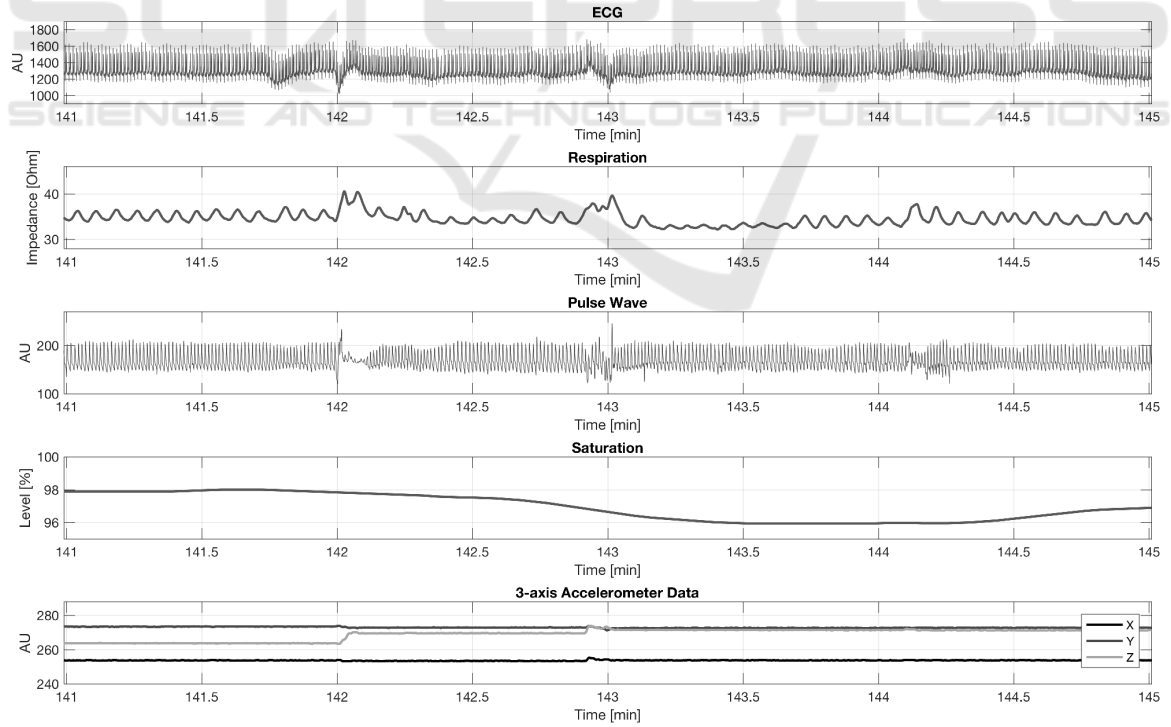

Figure 6: Sample *ECG*, smoothed impedance pneumography, pulse wave, smoothed blood saturation and accelerometry signals; with motion artifacts connected to the body position changes around 142 and 143 minutes of measurement; recorded during sleep using Pneumonitor 3.

ticular, whether it is feasible to measure quantitative parameters, particularly tidal volume and minute ventilation, as well as breathing phase timing, under "dynamic" conditions).

In our previous work, we confirmed that a calibration procedure is needed (particularly for various body positions) to measure quantitative parameters with accuracy comparable to the reference. However, we provided those results and consideration for static, "clear" conditions. They were free of possibly distorting maneuvers, such as irregular, shallow, intermittent breathing; quick movements; changes in electrode attachment (pressure of contact with the skin); or non-breathing-associated changes in the volume of the thorax, *etc.* These could influence the calibration coefficient in a way that would only allow one to state qualitative information about the depth of breathing.

One-lead *ECG* appears to provide results that do not allow clinical investigation and diagnosis, but do allow screening-like conclusions with greater comfort. Occurrences of the cardiac *IP* component may lead to the concept of removing the classic *ECG* measurement, however it seems, that cardiac *IP* component is not fully visible in each participant. In our opinion, if we are able to use the same electrodes to measure impedance and *ECG* signal, it will be worth having the redundant measurements for verification purposes.

Removing separated *ECG* input for combined one is prepared in order to improve the usability of Pneumonitor 3. It is also related with motion tracking. We decided to limit ourselves to a single inertial unit (including both accelerometer and gyroscope sensors) to maintain comfort and accuracy of detecting basic body positions and activity levels (Bouten et al., 1997; Ermes et al., 2008).

5 CONCLUSIONS

Prototypes of two devices intended for cardiorespiratory measurements in ambulatory conditions (with motion tracking, including activity and body position changes) were prepared and preliminarily evaluated.

The devices utilize impedance pneumography to calculate respiratory parameters (tidal volume, minute ventilation and respiratory rate) after calibration, *ECG* to record heart activity, and a combination of 3-axis accelerometer and gyroscope to track motion. Pneumonitor 3 also includes pulse oximetry providing saturation and pulse wave, and uses 5 electrodes setting with 2 *ECG* electrodes combined with receiving *IP* ones.

Pneumonitor 2 is designed for the environment physiology analyses (registering ventilation and cardiac functioning in subjects with obesity or nervous-muscle-related illnesses) and sports medicine (for ambulatory diagnostics, monitoring training, and determining exercise capacity). Pneumonitor 3 is intended mainly for sleep studies to monitor breathing disorders and the treatment progress.

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