Optimization of the Position of Single-Lead Wireless Sensor with Low Electrodes Separation Distance for ECG-Derived Respiration

Helena SVOBODOVA[1], *Erik VAVRINSKY*[1,2], *Daniela TURONOVA*[3], *Martin DONOVAL*[2,4], *Martin DARICEK*[2,4], *Peter TELEK*[2], *Martin KOPANI*[1]

[1]Institute of Medical Physics, Biophysics, Informatics and Telemedicine, Faculty of Medicine, Comenius University, Sasinkova 2, 813 72 Bratislava, Slovak Republic
[2]Institute of Electronics and Photonics, Faculty of Electrical Engineering and Information Technology, Slovak University of Technology, Ilkovicova 3, 812 19 Bratislava, Slovak Republic
[3]Department of Psychology, Faculty of Arts, Comenius University, Gondova 2, 81499 Bratislava, Slovak Republic
[4]NanoDesign ltd., Drotarska 19A, 81104 Bratislava, Slovak Republic

svobodovahs@gmail.com, erik.vavrinsky@stuba.sk, mosdan11@gmail.com, martin.donoval@stuba.sk, martin.daricek@stuba.sk, peter.telek@stuba.sk, martin.kopani@fmed.uniba.sk

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Abstract. *ECG - Derived Respiration (EDR) is a classical method for estimation of respiratory information from Electrocardiogram (ECG). This method normally uses flexible electrodes located at standard electrocardiography positions. This study introduces an alternative approach suitable for miniaturized sensors with low inter-electrode separation and electrodes fixed to the sensor encapsulation. Application of EDR amplitude algorithm on a single-lead wireless sensor system with optimized electrode positions shows results comparable with standard robust systems. The modified method can be applied in daily physiological monitoring, in sleep research studies or it can be implemented in smart clothes when standard respiration techniques are not suitable. In this paper we report optimization of the positions of miniaturized device for measurement of ECG and ECG-derived respiration.*

Keywords

ECG, EDR amplitude algorithm, miniaturized, respiratory frequency, wireless sensor.

1. Introduction

Commonly produced and commercially used medical, psychophysiological or intended devices for sport activities have been overwhelmingly designed to minimize additional artefacts. This situation partially results from past times, when analogue systems were used almost exclusively. As an advantage, the systems are easy to operate. However, useful additional information could be lost [1] and [2]. Nowadays, new digital systems are gaining ground. Technical progress facilitates the development of new measurement methods with possibility to obtain multiple physiological parameters from a single bio-signal [3], [4], [5], [6], [7], [8], [9] and [10]. The respiratory frequency is usually measured with spirometry, pneumography or plethysmography techniques. Upon these methods, the usage of cumbersome devices which may interfere with natural breathing, and which are often inapplicable in certain kinds of measurements such as daily monitoring at home, through stress testing, or during sleep research studies is not required [11], [12], [13], [14] and [15].

Several methods were proposed to derive the respiratory signal from the ECG using beat-to-beat morphologic variations. It is well known that the respiratory activity influences electrocardiographic measurements in various ways [16] and [17]. ECG-derived respiration method we have used, is evolved from chest movements and changes of the impedance distribution in human thorax that affects the amplitude of the QRS complex during the respiratory cycle [7] and [18]. We used this method derive EDR from the magnitude of QRS complex for our holter device. Breathing causes a rotation of the electrical axis of the heart which also affects the beat morphology. The heart movement is mostly affected by diaphragm, which moves the heart vertically and rotates it (Fig. 1) with respect to the body surface, i.e. towards the electrodes [19]. Therefore, the signal depends also on the type of respiration of each individual proband. Male and female probands commonly use different style of respiration. For male probands, so-called abdominal breathing is dominant; for female probands a costal respiration is typical [20] and [21].

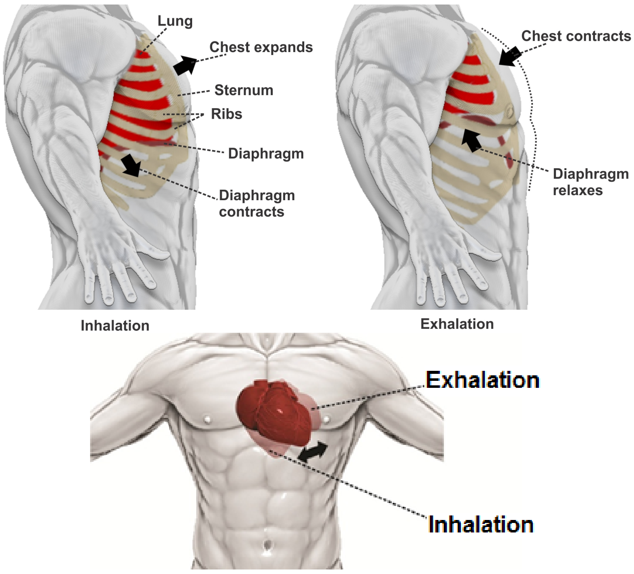


Fig. 1: Breathing process and diaphragm movement and changes in chest position during exhalation and inhalation.

Nowadays, continuous physiological monitoring systems (holters) and the setup with low number of miniature fixed electrodes is needed. For miniaturization, electrodes must have low separation, which is comparable with the heart displacement in thorax. This creates new challenges such as signal amplitude decrease and higher interference with the heart displacement. In other words, the sensor system is more sensitive to the electrodes placement and to the rotation on the body. Therefore, our aim is to develop standard and basic rules for wireless single lead ECG measurement. We want to determine appropriate location of our wireless Radio Frequency (RF) sensor system on the human body in order to obtain optimal quality of ECG and respiration frequency signal as well as comfortable wearing during long-term monitoring [22] and [23]. Recently, various methods were proposed to extract the surrogate EDR signal [4], [7], [16], [22], [23], [24], [25] and [26] based on the amplitude of QRS complex, R-wave duration and QRS complex area, T-wave amplitude or T-wave area. In our single-lead system with "hard fixed" electrodes we have chosen the amplitude of QRS complex method as suitable to use.

In this article, the modification of standard EDR method for low inter-electrode separation and electrodes fixed to the sensor encapsulation is demonstrated. The overall goal is to develop a universal multi-sensor for daily use, 24-hour monitoring, with multi-electrode array applicable in e-health monitoring and in the scientific research.

2. Material and Methods

2.1. Sensor System

The proposed sensor system (Fig. 2, Tab. 1) used in this study is composed of autonomous sensing units designed for wireless "real-time" biopotential (ECG, EMG) measurement and evaluation, with the possibility of a simultaneous monitoring of different parts of the body. PC communication is provided by one receiver module (USB dongle), which can simultaneously communicate with 4 to 6 sensing units. The system has been designed for daily monitoring of health in rehabilitation centres as well as for remote healthcare at home, since it is possible to continuously monitor the cardiac and muscle activity. The sensing unit is based on precise instrumentation amplifier designed for bipolar detection arrangement. The gain of the amplifier can be changed in the range from 40 to 80 dB, where the device also includes a 10× preamplifier. Measuring module includes 24-bit sigma/delta AD converter and 32-bit 16 MHz processor. Sampling frequency is programmable in the range from 5 to 2000 SPS (samples per second). The gain and sampling frequency can be set online. The determination of physical parameters such as acceleration and position is provided by a built-in 3D accelerometer and magnetometer. Dimensions of the sensor electrodes are 1×2 cm and the electrodes are in the standard "EMG" distance, which is 2 cm.

Fig. 2: Sensor system (holter). Up - PCB design of sensing unit, Left down - Sensing unit, Right down - Receiver module.

Tab. 1: Technical parameters of sensor system.

Input range	10 mV pp (can be switched to 100 mV pp)
Input impedance	5 MΩ, 100 pF
Noise	33 nV·Hz$^{-1/2}$
Programmable amplifier	40–80 dB
ADC resolution	24-bit, no data missing
Sample rate	5–2000 SPS
Acceleration sensor	3D, 10/12/14 bit, 100 SPS
MCU	32 bit 16 MHz
Synchronization error	$\leq$ 30 µs
Connectivity	ISM band 2.4 GHz, standard IEEE 802.15.4
Measuring network	4–6 probes per one dongle
Transmission range	10 m range at interior, 30 m range at exterior
Supply voltage	1× Li-Pol 120 mAh, 4 h stamina, charging 20 min
Temperature range	0–70 °C
Electrodes	10 × 20 mm, Au galvanic plated on Cu
Connector	Micro USB
Weight	20 g
Dimensions	37 × 38 × 16 mm
Next features: Automatized network administration RGB LED signalization, magnetometer, automatic switch on/off, on-line configurable parameters: sampling rate digital filter and PGA, colour of RGB	

2.2. Chest Mapping

The experiment was composed of three consecutive steps. Firstly, a detailed mapping of signal quality considering the location of the sensor on the thorax was realized. It is due to the fact that sensor system has electrodes fixed to the encapsulation and the electrodes have significantly lower separation than wire electrodes used in standard holter devices, where the distance between electrodes varies between 5 and 20 cm. The conventional EDR method is difficult to apply for this electrode setup because the heart displacement is comparable to the electrodes separation. The dynamic motion of the heart in thorax is a very critical parameter because this motion is greater than 2 cm and the number of inaccuracies of conventional EDR method is increasing. This experiment was realized on 8 probands (4 males, 4 females); 600 measurements in total. The probands were in sitting position. The location of the sensor was intentionally chosen non-professionally since the system was designed for home use. Measurements were done at points in square grid with reference points: central vertical line, nipple, and bottom edge line of the pectoral muscle. The "chest mapping" was done on the thorax area close to heart at 15 different measurement points. On each position, we measured signals at 5 different electrode rotations (0°, 45°, 90°, 135°, and 180° with controlling point, 0° = 180°) as illustrated in Fig. 3. Each of 600 measurements consisted of 3 periods, each of 10 seconds duration (10 seconds of normal respiration followed by 10 seconds of

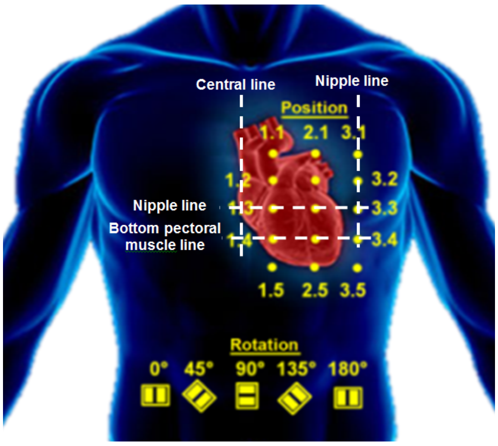


Fig. 3: The measuring positions of the sensor with reference measuring lines. Each yellow spot shows area for electrode fixing. We used sensor system (holter) in 0°, 45, 90°, 135° and 180° (controlling point = 0°) electrodes rotations.

breath-hold inhalation and 10 seconds of breath-hold exhalation).

ECG data were measured at sampling rate 1000 SPS. For EDR evaluation, the measured raw (unfiltered) data were filtered using software band-pass filters with 3 Hz low cut-off frequency and 100 Hz upper cut-off frequency. The QRS amplitude of the filtered signal, which corresponds to the respiratory information, was calculated for each heartbeat. To determine the suitable sensor position and rotation we decided to use subjective criteria. Five professionals with years of experience were asked to evaluate the quality of measured signal by giving each from 600 measured signals points between 0 (the worst) and 5 (the best). The criteria were to clearly distinguish P-R, QRS, and S-T interval of ECG waveform and at the same time, the amplitude fluctuation due to respiration.

2.3. Influence of Daily Activities on the Measured Signal

The second step included laboratory testing of the measurement positions identified in the first experiment and a broader group of probands who were undergoing a physical test. In this study, 28 probands were measured (mean age = 23.5 years; SD = 1.41; 13 males; 15 females) at physical and psychological (Dual N-back) tasks. Probands were asked to report whether they use any use of medication or suffered from a chronic disease. The whole task took about 30 minutes and the defined activities where physically set to simulate real life conditions. The experiment started with a 10 minutes relaxation phase. Participants were instructed to lie down on bed and relax. Thereafter, participants were facilitated into a neutral sitting posture for 5 min, on a standard office chair. In the next phase participants were instructed to perform a Dual N-back task. The N-back task is a continuous performance task used to assess individuals' levels of processing capacity, and individual differences in working memory capacity. We used a dual N-back task at three levels of difficulty, with auditory and visual-nonverbal material, presented simultaneously. The task requires on-line monitoring, updating, and manipulation of remembered information and therefore it is assumed to place great demands on a number of key processes within working memory [24] and [27]. After this cognitive task, participants were asked to stand up against the wall for 3 minutes, to perform 20 squats, and again stand up against the wall for 3 minutes. We measured each proband in different human postures and we registered the score at the optimal positions selected in the first step of study.

2.4. Comparison with Reference System in Daily Use

Two final measurement points with highest score were obtained from second experiment and they were used in last part of our work. In order to evaluate quality and reliability of the electrode system with low electrode separation, the derived respiratory information was compared with suitable reference measurement of similar nature by twin chest belt and standard single chest belt in daily monitoring (12 hours) on 2 volunteers (1 male, 1 female). We used 2 belts for reference system; one chest belt monitored abdominal breathing while the second one recorded thoracic breathing. In contrast to classical chest belt, the twin chest belt system can monitor different styles of respiration very reliably even in complicated situations [9]. Both belts were connected to recording device (Nexus-10 MKII [28]) where the signal was averaged and used as a reference. The sample rate of reference method was set to 32 SPS. The sensor system with low electrode separation was placed to two positions specified in previous step of study. Sample rate was set to 1000 SPS. Both compared systems were wirelessly connected to one computer and the synchronization was achieved by using manual triggering keys. The probands were long-term monitored in their typical daily situations. Specifically, it was a typical office work like sitting, sleeping and physical activities, such as cycling, gait, running, and squatting.

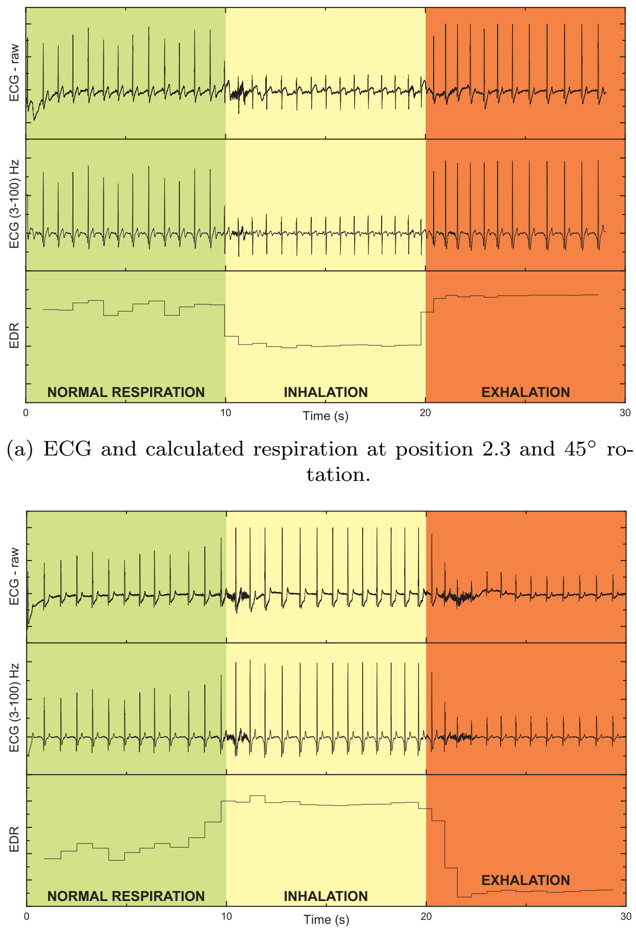
3. Results

3.1. Chest Mapping

The results of signal quality testing with respect to the electrodes position on the chest are summarized in Tab. 2. The chest mapping data showed optimal position (the averaged points from all rotations) for male probands - 3.2/45°, female probands - 2.1/0°, and optimal position calculated for both genders - 3.1/0°. Subsequently, we observed and calculated all suitable positions for both genders together and comfortable wearing, i.e. that person is not limited by our holter. The comfortable position allows performing of general activities without limitations and the device does not damage clothes. The best positions were observed on the lower part of chest, at horizontal measuring lines 4 and 5. Two positions with the highest score were

Tab. 2: Optimal measuring positions for male, female, both genders together and positions for comfortable wearing with the best point score.

Proband	Optimal position-rotation	Aver. points score
Male	3.2/45°	4.6 ± 0.7
Female	2.1/0°	4.9 ± 0.3
Male + Female	3.1/0°	4.4 ± 0.9
Male + Female (comfortable wearing)	2.4/135°	3.5 ± 1.7
	3.4/0°	4 ± 1.5



(a) ECG and calculated respiration at position 2.3 and 45° rotation.

(b) ECG and calculated EDR at position 3.5 and 45° rotation.

Fig. 4: Typical measured ECG, ECG after band-pass filtration and calculated EDR signal in time.

chosen - 2.4/135° and 3.4/0° (Tab. 2). Examples of measured data are shown in Fig. 4. Figure 4(a) and Fig. 4(b) depict typical signals from upper part of thorax (position 2.3 / rotation 45°) and lower part of thorax (position 3.5 / rotation 45°), respectively. Herein, it should be mentioned that EDR signal flips over the horizontal axis between top and down part of thorax considering to breathing phase. This effect appears typically in thin line around 4 and 5 row of measuring points and is for each proband because of different physique and respiration style of an individual. It can be overcome with careful vertical shift of the electrodes on the chest. Even though there are several "blind positions" where the signal is insensitive to respiration, however, the line comprising these positions is very thin.

3.2. Influence of Daily Activities on Measured Signal

Tab. 3: EDR signal quality in simulated daily activities - complex testing (average scores at the best positions and decrease in signal quality compared to chest mapping of the sitting probands). The positions with the best scores for daily usage of our device are 2.4/135° and 3.1/0°.

Proband	Electrod. position -rotation	Aver. points score	Changes in signal quality
Male	3.2/45°	3.3 ± 1.7	-28 %
Female	2.1/0°	3.5 ± 1.5	-29 %
Male + Female	3.1/0°	3.7 ± 1 / 3.8 ± 1.1	-16 %/-14 %
Male + Female (comfortable wearing	2.4/135°	3.6 ± 1.3 / 2.9 ± 1.1	+3 %/-17 %
	3.4/0°	2.8 ± 1.5 / 1.9 ± 1.1	-30 %/-53 %

The chest mapping provided five optimal holter positions, which were subsequently tested in simulated daily activities. Figure 5 illustrates a typical measurement of one proband during different human postures. We evaluated the score for daily usage at the 5 optimal positions (according to the same criteria as in Subsec. 3.1.). The EDR score decreased in comparison with the first experiment "Chest mapping", when the probands were measured only in sitting positions. Scores are summarized in Tab. 3. To verify the EDR method accuracy, all studied positions were evaluated separately for each gender. The decrease in signal quality at the best scored positions in comparison to chest mapping (Tab. 2) for daily usage was measured. The result is 28 % decrease for male probands, 29 % decrease for female probands, 15 % for both genders (16 % for male, 14 % for female probands) and 7 % (even 3 % increase for male and 17 % decrease for female probands) and 41 % (30 % for male and 53 % for female probands) for comfortable wearing for both genders. Based on our results, we continued in our measurements at 2 tested positions with the highest score and lowest decrease in signal quality (3.1/0° and 2.4/135°), which appeared to be most suitable for both genders.

3.3. Comparison with Reference System in Daily Use

For final comparison with reference system, two positions with highest score from second experiment were chosen: 3.1/0° and 2.4/135°. The results show that the proposed electrode system has a reliable performance comparable with the reference system. Typical results acquired from one male proband measured in sensor position 2.4/135° are shown in Fig. 6. The average differences between respiratory frequency acquired from EDR and reference device (twin chest belt) in everyday life and long-term monitoring is 0.56 % at 2.4/135° position and 0.33 % at 3.1/0° position for sitting, 1.81 % a and 0.79 % for lying, 1.01 % and 4.22 % for squatting, 10.4 % and 15.3 % at 3.1/0° for cycling and 42.27 % a and 44.85 % for running (Tab. 4). This method is not appropriate for measurement of activities, when the heart motion in thorax is very fast and random. The worst results of EDR method in comparison to the reference method were obtained during fast running of our probands; the difference in respiratory frequencies is 42.27 % compared to the use of single chest belt, when the difference is only 6.97 % at 2.4/135° position and the difference in respiratory frequencies is 44.85 % compared to the use of single chest belt, when the difference is only 5.75 % at 3.1/135° position.

Tab. 4: The differences between respiratory frequencies acquired from EDR system, single belt and reference device in daily use.

Difference in respiration frequency between EDR system, single belt and reference method		
Position 2.4/135°		
Activity	**EDR/Reference Respiratory frequency**	**Belt/Reference Respiratory frequency**
Sitting	0.56 %	3.55%
Lying	1.81 %	0.08%
Cycling	10.4 %	8.21%
Squatting	1.01 %	0.56%
Running	42.27 %	6.97%
Position 3.1/0°		
Sitting	0.33 %	3.35 %
Lying	0.79 %	0.17 %
Cycling	15.3 %	8.89 %
Squatting	4.22 %	0.78 %
Running	44.85 %	5.75 %

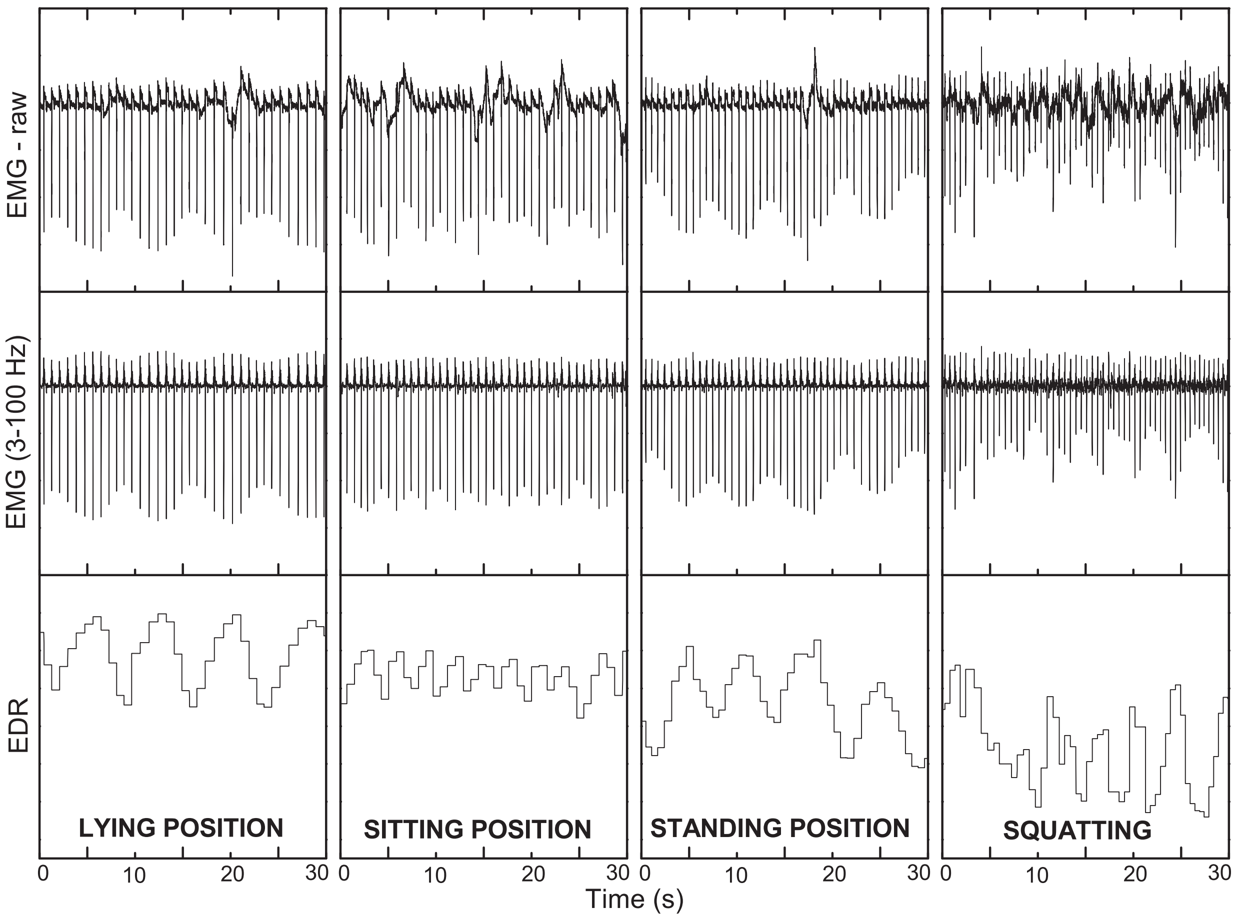


Fig. 5: Typical ECG and corresponding EDR signal at different postures continuously measured in laboratory on one proband.

4. Discussion and Conclusion

In this article, we demonstrated application and optimization of the amplitude ECG-derived algorithm for estimation of respiratory frequency using a single-lead wireless sensor with constant electrode separation of 2 cm. Since the electrode separation is comparable with the heart displacement due to motion of thoracic diaphragm, the electrode position is crucial for the reliable EDR measurement. The mapping at chest by variation of probe position/rotation provided an optimal setup to achieve signal quality on the level of reference twin chest belt method. Our results show that 2.4/135° position is optimal for simultaneously monitoring of ECG and EDR for both genders in daily usage. The reason could be the close position of the heart and the electrodes. The movement of the heart (heart apex) is greatest at position between the electrodes of the sensor system, which is showed in Fig. 1 down. The position 2.4/135° is approximately vertical to electrical axis of the heart. We recommend the use of this position for other measurements and for usage of our device. Further sensor testing at optimal positions revealed reliable respiration measurements for various daily physical activities. From all achieved results, the conclusion can be made that used EDR method offers very stable and solid results. Limitations occur only at specific physical activities, when the heart position in thorax is very unstable. Therefore, the suggested electrode setup is suitable in daily monitoring, if standard respiration monitoring techniques are inappropriate for some reason. For common daily activities, in some cases even better results were achieved by the proposed sensor system in comparison with standard single chest belts in daily use. Better results were observed for EDR under lower physical activity and during talking. Excellent results were obtained during sitting, lying and squatting of our probands. Acceptable results were measured during cycling, but specific physical activities such as running of our probands showed limitations of our holter device. The standard thoracic chest belt showed better results under higher physical activity but some activities, like talking, affect the results more in comparison with our sensor system. During running, the EDR algorithm is inefficient due to the heart is in thorax hopping. The miniaturized setup, where electrodes are fixed to the encapsulation of the sensor, bring the possibility of its combination with other sensors such as electro-dermal activity monitoring or reflectance stereo pulse oximetry systems. As

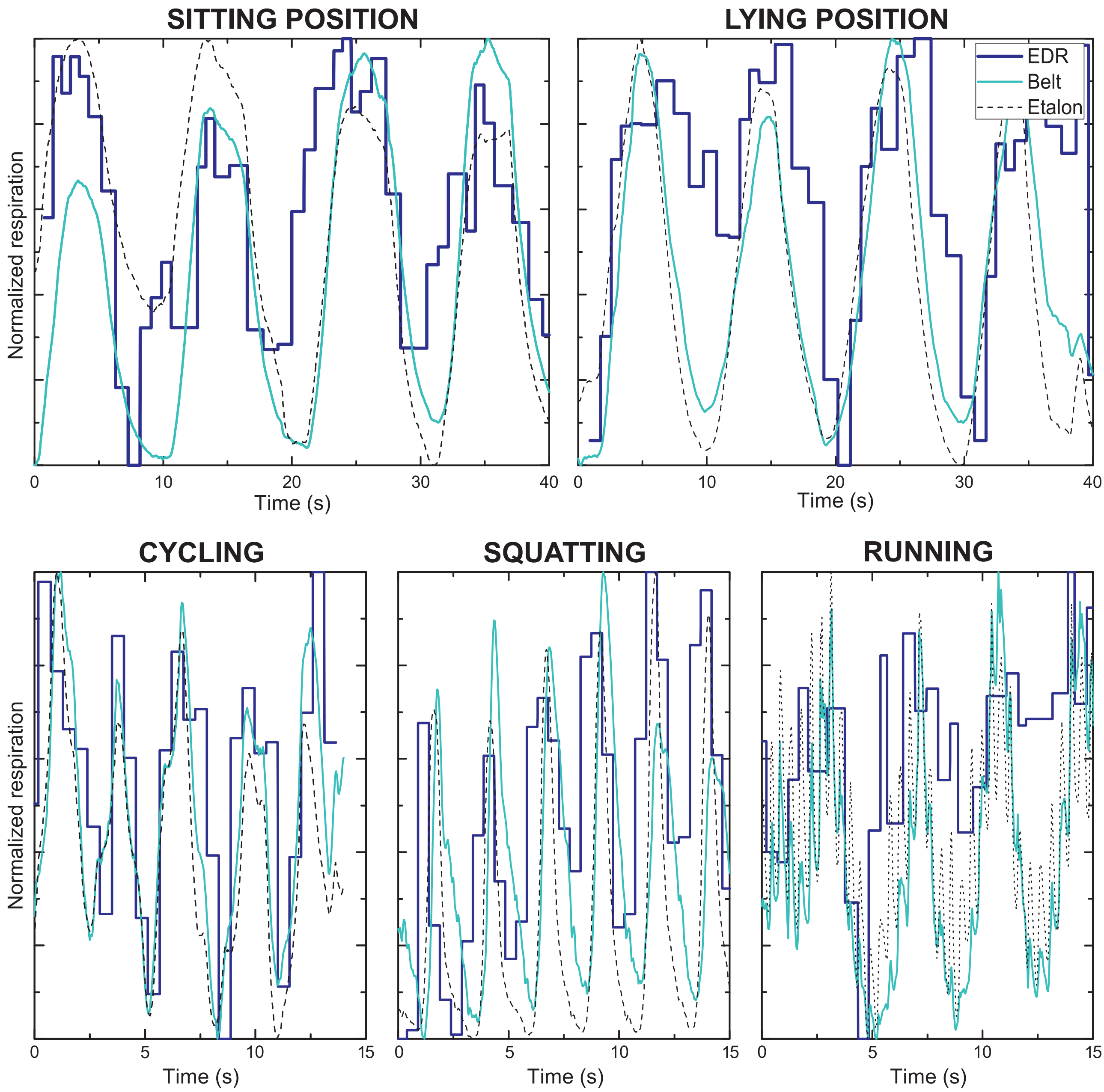


Fig. 6: Practical comparison of our EDR system in sensor position 2.4/135° with reference systems measured simultaneously on one proband during 12 hours.

a result, the proposed ECG/EDR electrode setup is suitable for compact multipurpose sensor system and opens many new possibilities in the reliable daily monitoring of human physiological parameters. The results achieved from the chest mapping will support the realization of smart clothing.

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About Authors

Helena SVOBODOVA obtained her Master's degree in the field of molecular biology at Faculty of Natural Science, Comenius University, in 2014. In present, she is a Ph.D. student at the department of biophysics, Faculty of Medicine. Her research interests include metabolism of iron, neurodegenerative diseases, electroencephalo-graphy, electromyography, electrocardiography, sensor systems and bio-monitoring devices. In 2017, she worked at University of Eastern Finland on models of Alzheimer disease.

Erik VAVRINSKY received his Masters and Ph.D. degree from the Faculty of Electrical Engineering and Information Technology, Slovak University of Technology in years 2002 and 2005, respectively. He has carried out postgraduate studies in area of thin-film microsensors and electrochemical biosensors. In present, he participates at the University education process as a lecturer. His current research interests include design, characterization and testing of sensor systems and bio-monitoring devices. Dominant part of his multidisciplinary research is focused on human physiology.

Daniela TURONOVA received her Master's

degree (2008) in Psychology, and Ph.D. (2012) in Clinical Psychology from Comenius University in Bratislava. She is currently an Assistant at Department of Psychology, Faculty of Arts, Comenius University in Bratislava. Her research activities are in the field of Cognitive Psychophysiology and Psychobiology.

Martin DONOVAL received his Master's and Ph.D. degrees in Electronics from Slovak University of Technology in Bratislava. He has carried out postgraduate studies in area of IC design. He participated in the establishment of the Department of electronic systems with focus on reliability testing, design of smart electronic systems and off-chip innovative solutions. Currently, he is responsible for projects particularly in a field of bio-electronics. As a member of the team he is involved in development of new prototypes and dedicated integrable electronics and reliability testing.

Martin DARICEK studied at Slovak University of Technology in Bratislava where he received his Ph.D. degree in field of device and IC characterization Currently he focuses on off-chip innovative measuring, testing and bio-monitoring devices design, characterization, development and optimization. He is as well responsible for preparation of low-level firmware schemes. Besides this is he currently responsible for R&D projects management. In present, he participates at the University education process as a lecturer.

Peter TELEK studied at Slovak University of Technology in Bratislava where he in year 2013 received his Ph.D. degree in field of optoelectronics. Name of the thesis was "Properties of light emitting diodes for spintronics application".

Martin KOPANI head of Institute of Medical Physics, Biophysics, Informatics and Telemedicine. His research activities focus on application of various physical methods in study of cells and tissues, mainly in terms of the presence, localization, distribution, and structure of some chemical elements.