Comparative Evaluation of Susceptibility to Motion Artifact in Different Wearable Systems for Monitoring Respiratory Rate

Antonio Lanatà, Enzo Pasquale Scilingo, Elena Nardini, Giannicola Loriga, Rita Paradiso, and Danilo De-Rossi

***Abstract*—The purpose of this study is to comparatively evaluate the performance of different wearable systems based on indirect breathing monitoring in terms of susceptibility to motion artifacts. These performances are compared with direct respiratory measurements using a spirometer, which is accurate, reliable, and less sensitive to movement artifacts, but cannot be integrated into truly wearable form. Experiments were carried out on four indirect methods implemented into wearable systems, inductive plethysmography, impedance plethysmography, piezoresistive pneumography, and piezoelectric pneumography, to ascertain the performance of each of them in terms of noise due to movement artifacts, as well as to study the effects of different movements or gestures during each test. A group of volunteers was asked to wear all of the breath monitoring systems simultaneously along with the face mask of the spirometer while carrying out four physical exercises in a gym under controlled conditions. Data are analyzed in the time and frequency domain to estimate the frequency respiration from each wearable system and compare it with those of the spirometer. Results confirmed that all the wearable systems are somehow affected by movement artifacts, but statistical investigation showed that for most of the physical exercises, three out of four, piezoelectric pneumography provided best performance in terms of robustness and reduced susceptibility to movement artifacts.**

***Index Terms*—Plethysmography, pneumography, respiratory rate monitoring, spirometer, wearable systems.**

I. Introduction

NEW HEALTH care practices are increasing their focus on personalized health care by using devices that continuously monitor vital and behavioral signs, helped by advances in sensor and communication technologies along with data processing [1]. These technologies are currently being used to overcome the challenges that continuous monitoring is present in the development of wearable health monitoring and diagnostic systems, often implemented using electronic textiles [2]. Continuous monitoring poses many challenges for all types of physiological signs, but the simplest, paradoxically being the most challenging vital sign to accurately record, is respiratory activity due to the fact that the signals are affected by movement artifacts and filtering, or feature recognition algorithms are not very effective. Monitoring of respiratory activity involves the collection of data on the amount and the rate at which air passes into and out of the lungs over a given period of time. In the literature, there are several methods to do this, both directly, by measuring the amount of air exchanged during the respiration activity, and indirectly, by measuring parameters physically correlated to breathing, such as changes in thorax circumference and/or cross section, or transthoracic impedance. Direct methods use a spyrometer that measures directly the airflow in the lung exchanged during inspiration and expiration. However, this device cannot be implemented in wearable systems because it employs a mouthpiece, which could interfere with the freedom of movements, disrupting the normal breathing pattern during measurement, thus causing discomfort for the user. Indirect methods exploit displacements of the lung that are transmitted to the thorax wall and *vice versa*, and therefore, measurements of chest–abdominal surface movements can be used to estimate lung volume variation. A number of devices have been used in the past to measure rib cage and abdominal motion, including mercury in rubber strain gauges [3], linear differential transducers [4], magnetometers [5], and optical techniques [6]. Among these systems, the first three have not been extensively used in research and clinical practice. Optical techniques have been introduced several years ago, but the use in research and clinical practice has been consistent only after 1994 when an optoelectronic 3-D motion analysis system (Elite system [7]) was developed; this system presents a considerable drawback in which it is cumbersome, extremely expensive, and can only be used in research environments or in laboratory applications. Indirect techniques that can be implemented in wearable systems are respiratory inductive plethysmography [8], impedance plethysmography [9], piezoresistive pneumography [10], and/or piezoelectric pneumography. These systems are minimally invasive and do not interfere with physical activity. Wearable property implies using textile substrate or similar, which can be easily affected by movement artifacts due, for example, to slipping or movement during activity. This issue is addressed in [11]. Herein lies the challenge undertaken in this study to

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accurately monitor respiratory activity when a subject is doing physical exercise or activity. The four indirect respiratory monitoring methods mentioned previously were tested during the collection of respiratory data and compared to the results obtained by the spyrometer. In this study, we performed a set of experiments in order to evaluate how controlled movements can affect performance in terms of respiration rate recognition. Four wearable systems based on the four aforementioned methodologies were tested in comparison with a spyrometer on a group of ten volunteers. A specific experimental protocol implying a set of predetermined physical exercises was set out. The systems are worn simultaneously and signals are processed in the time and frequency domain, and then statistically treated. The goal is to identify which wearable system is more robust to motion artifacts, thus providing a statistical assessment of the agreement between each system and the spyrometer in terms of respiration rate recognition.

II. Materials and Methods

A. Instrumentation

The four different methodologies to monitor respiration activity, inductive plethysmography, impedance plethysmography, piezoresistive pneumography, and piezoelectric pneumography, which can be implemented into wearable and comfortable systems to indirectly monitor respiratory activity, were compared in terms of performance with respect to a spirometer. In inductive plethysmography, piezoresistive pneumography, and piezoelectric pneumography, only one sensing belt wrapped around the thorax was used regardless of the abdomen. This choice is justified by the fact that, normally, the inspiratory thoracic and abdominal expansion is almost synchronous; therefore, no significant differences can be detected. Only when there is an increased upper airway resistance or obstruction, the thorax moves inward and the abdomen outward during inspiration, and in expiration, this pattern is reversed (paradoxical rib cage and abdominal movements). This represents a complete phase shift of movements from the normal pattern, but if the upper airway is only partially obstructed, there may be only a change in the phase angle and timing of the movements of the thorax and abdomen [12]. However, we experimentally validated the synchronism between chest and abdomen signals in rest configuration as well as in physical activities. An experimental protocol was set up where a group of ten volunteers was asked to perform specific physical exercises when the respiratory activity was recorded. In the following, a description of the techniques as well as the instrumentation used during the experimental sessions are reported.

1) Spirometer: Spirometers provide a direct, noninvasive measurement of the airflow through the lungs, and are commonly accepted as a gold standard in clinical practice for respiratory monitoring. Spirometers are usually based on pressure transducers, which measure a signal proportional to the difference of pressure between the two sides of a membrane that causes a pressure drop proportional to the airflow. During the experiments, an airflow transducer $SS11LA$ along with the $MP100$ data acquisition system by $Biopac$ System, Inc. was used [13]. Hereafter, we will refer to the spirometer as *Biopac*. The $SS11LA$ is intended for human use, and has a removable head for sterilization and replacement. For easy execution of the tests, and to guarantee free movement for the subjects, an apposite face mask fixed by a head strap (accessories of the $Biopac$ System) was used throughout all the experiments. In Fig. 1, the acquisition system with the face mask is shown. The face mask comprises a mouthpiece to collect the air that is inhaled and exhaled. No amplitude calibration procedures were carried out before the experiments, considering that the aim of the study was the correct identification of the respiratory rate by analyzing the signal waveform through a suitable algorithm; therefore, measuring the airflow expressed in liters per second is beyond the purpose of this paper. The MP100 device allows for setting the sampling rate, the amplification, and the cutoff frequency of the filters during acquisition. The data were acquired at a sampling frequency of 100 Hz, and a second-order low-pass filter with a cutoff frequency of 60 Hz and a Q value of 0.7 was applied.

Fig. 1. Biopac system used as spirometer along with the facemask.

2) Inductive Plethysmography: The inductive plethysmography method for breathing monitoring consists of two conductive wires placed around the thorax and the abdomen in order to detect the cross-sectional area changes of the rib cage and the abdomen region during the respiratory cycles. The conductive wires can be considered as a coil and are used to modulate the output frequency of a sinewave current produced by an electric oscillator circuit. As a matter of fact, the sinewave current generates a magnetic field, and the cross-sectional area changes due to the respiratory movements of the rib cage and the abdomen that determine a variation of the magnetic field flow through the coils. This change in flow causes a variation of the self-inductance of each coil that modulates the output frequency of the sinusoidal oscillator. This relationship allows to monitor the respiratory activity by detecting the frequency change in the oscillator output signal. The experiments were carried out by using Vitaport cardiopulmonary screener (CPS) by *Temec* (see Fig. 2) [14], hereafter referred to as *Temec*. This commercial system comprises two inductive belts that wrap around the thorax and the abdomen, respectively, and the received signals are conveyed to an electronic processing unit. For the sake of uniformity with the other instruments, during the experiments performed during this study, only the thorax signal was

Fig. 2. Vitaport CPS system along with the two belts containing the inductive coils.

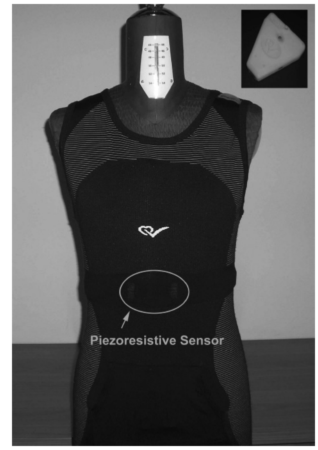

Fig. 4. MyHeart shirt with the piezoresistive sensor highlighted by a circle. In the upper right corner, the PPU is shown.

Fig. 3. Wealthy shirt. E1 and E4 are the electrodes used to inject current and E2 and E3 are used to acquire the voltage drop. Dividing the voltage by the current, we can obtain the transthoracic electric impedance. In the upper right corner, the PPU is shown.

considered in the analysis. The Vitaport CPS device is provided with 15 channels for the acquisition of different physiological parameters and vital signs. The respiration signal was acquired with a sampling rate of 32 Hz.

3) Impedance Plethysmography: This technique consists of injecting a high-frequency and low-amplitude current through a pair of electrodes placed on the thorax and measuring the transthoracic electrical impedance changes [15]. These electrodes are made of fabric and integrated into the Wealthy shirt, a wearable system developed by Smartex s.r.l. within the frame of the European cofunded Wealthy Project, IST-2001-37778, hereafter referred to as Wealthy. As a matter of fact, there is a relationship between the flow of air through the lungs and the impedance change of the thorax. The measurements can be carried out by using either two or four electrode configurations. During the experiments, the Wealthy shirt (see Fig. 3) was used, as it was able to perform impedance pneumography [16] by using four textile electrodes integrated in a one-step process in the fabric of the shirt by means of a flat knitting machine [17]. The data were acquired by using a suitably designed portable patient unit (PPU). The PPU is a portable device for the acquisition of different physiological signals based on a $Bluetooth$ module for data transmission. As far as the respiratory activity monitoring is concerned, a high-frequency (50 kHz) and low-amplitude current ($500\mu A$) is injected by the outer electrodes placed on the chest, whereas the inner electrodes are used to monitor the impedance changes by measuring the voltage drop variations between them. The respiration signal was sampled at 15.625 Hz.

4) Pneumography Based on Piezoresistive Textile Sensor: Piezoresistive pneumography is carried out by textile sensors that monitor the cross-sectional variations of the rib cage. These sensors are integrated into the "Take Care" sensorized shirt developed by Smartex s.r.l. within the frame of the European cofunded My Heart project, IST-507816, hereinafter referred to as MyHeart. The piezoresistive sensor changes its electrical resistance if stretched or shortened, and is sensitive to the thoracic circumference variations that occur during respiration. During the experiments, the Take Care sensorized shirt was used (see Fig. 4). The textile piezoresistive sensor was integrated in a one-step process in the fabric by means of a circular knitting machine [18]. The data were acquired by a suitable portable device based on a $Bluetooth$ module for data transmitting. The respiration signal was acquired with a sampling rate of 25 Hz.

5) Plethysmography Based on Piezoelectric Sensor: This method is based on a piezoelectric coaxial cable sensor, produced by measurement specialties (MSI), whose sensing element is a piezoelectric polymer p(VDF-TrFE) [19]. It was simply fastened around the thorax during the experiments carried out during this study, thus monitoring the thorax circumference variations during the respiratory activity. This device is hereafter referred to as piezoelectric device or PED. From datasheet, this sensor shows a linear behavior of the peak charge against stress. In particular, this linearity is held with stress variation from 0.5 to 3.5 kg/mm^2, whereupon a saturation phenomenon occurs; thus, when the cable is compressed or stretched, a proportional charge or voltage is generated. During the experiments, a belt containing a piezoelectric sensor in wire form was used (see Fig. 5), with a diameter of 1.07 mm, adjustable length of about

Fig. 5. Belt containing the piezoelectric sensor.

1 m, and a capacitance of about 200 pF/m. The analog front end of the acquisition device converts the charge variation of the piezoelectric sensor by means of a charge amplifier. An analog passband filter was implemented around the breathing rate ($f_L = 0.014$ Hz and $f_H = 0.5$ Hz). The acquired data are transmitted in real time to a laptop by means of a *ZigBee* short-range module. A friendly user interface developed in *Labview* software was used for data acquisition. The signal was sampled at 100 Hz. Further work is being developed to integrate this sensor into a wearable system. Some preliminary attempts have been done within the Proetex project, FP6-2004-IST-4-026987.

B. Testing Protocol

Ten volunteers participated in the experiment. In this study, only male subjects were involved in order to avoid interference by various breast sizes. Their ages ranged from 22 to 34 years, with an average of 28, and all had normal health conditions. For each subject, changes in transthoracic electrical impedance, respiratory inductance pletysmography, ribcage circumference, and airflow measurements are simultaneously recorded during a suitable protocol of physical activities. Each subject repeated the same activity four times. It is worthwhile noting that each system is electrically insulated and does not interfere with the others. Nevertheless, probably, mechanical overlap of the systems can slightly affect the accuracy in terms of amplitude, i.e., the thorax circumference measurement, but it is surely irrelevant for respiration rate detection, which is the goal of this paper. This does not impinge on the result validity. In the airflow measurement, subjects breathed through a mask connected to a Biopac central unit by a flexible tube; ribcage circumference variation measurement and inductive pneumography were performed by means of two sensing belts wrapped around the thorax all together at the level of the tip of the sternum; transthoracic electrical impedance measurements were obtained by means of textile skin contact electrodes placed along the nipple line. These electrodes exhibit low baseline impedance, good skin adhesion, good stability, and high flexibility. A hydrogel membrane was used to couple the electrodes to the skin. A preliminary test was carried out in order to verify if the spirometer is indeed less sensitive to movement artifacts. Subjects wore all the breath monitoring systems simultaneously in addition to the face mask of the spirometer. They were asked to do a specific AMNB exercise. It consists of raising arms at end exhalation and touching hands to top of head, returning arms to side, and repeating it five times. Further experiments are structured into four physical exercises done in a gym, each of 2 min in duration alternating 5 min of normal breathing. These exercises are walking, running, cycling, and elliptical trainer. The starting condition is meant to be normal breathing in a standing up position. In walking and running activities, the subject is asked to walk and run, respectively, on a treadmill at constant speed and constant inclination. During the cycling activity, the subject is asked to ride on a bicycle for 2 min, trying to keep the speed as constant as possible while maintaining the resistance and effort at a fixed level. During the elliptic activity, the volunteer is asked to use an elliptic trainer that is an exercise machine to induce specific and controlled upper and lower limb movements. Table I summarizes these exercises.

TABLE I
PHYSICAL EXERCISE PARAMETERS

Type of exercise	Support machines	Characteristics
Basal	—-	Breath quietly standing up
Walk	LifeFitness Treadmill	Speed: $5.5 Km/h$ Inclination: 8%
Run	LifeFitness Treadmill	Speed: $7 Km/h$ Inclination: 12%
Cycle	Technogym Bicycle	RPM 90 $6/12$ difficulty level
Elliptic	Technogym	—

C. Data Analysis

The signals coming from the four systems, as well as from the spirometer, are acquired and stored to be then processed offline. Signals were resampled to facilitate processing and then selectively filtered in order to increase SNR to reject frequency components outside the useful bandwidth. Next, two parallel algorithms were implemented providing both frequency- and time-domain analyses [20]. Finally, a statistical analysis is performed in order to generate statistical parameters showing agreement with the spirometer in terms of breathing rate identification, with a specified significance.

Hereafter, a detailed description of each aforementioned step is reported.

1) Resampling: Every instrument provides a signal output at different sampling frequencies: Biopac at 100 Hz, PED at 100 Hz, Vitaport at 32 Hz, Wealthy at 15.625 Hz, and MyHeart at 25 Hz. It implies a different signal resolution within the same time interval provided by each device. As the sampling rate of some of the devices cannot be modified, being an intrinsic manufacturing feature, a resampling method was used to uniform sampling frequency at 100 Hz. For undersampled signals with respect to 100 Hz, a resampling method introduces an estimation of the missing samples by means of cubic spline interpolating function. This strategy allows to compare signals with the same sample number over the same time interval. In addition, the spline interpolation uses a special type of piecewise polynomial as an interpolant and introduces a small error, avoiding the problem of Runge's phenomenon [21]. It is a problem that occurs when using polynomial interpolation with polynomials of high degree. This shows that high-degree polynomial interpolation at equidistant points can be troublesome.

2) Filtering: To improve the SNR, a filtering operation was carried out. As respiratory frequencies fall into a limited bandwidth, a high-selective passband filter was implemented in order to extract them. All the signals are processed by this filter. Filter is digitally implemented through a passband IIR Butterworth of 10th degree, which is maximally flat in the passband zone. Cutoff frequencies comprised between 0.15 and 0.5 Hz.

3) Frequency- and Time-Domain Analyses: An analysis in the time domain as well as in the frequency domain was performed. Time-domain processing is used to demonstrate that spirometer is only slightly affected by movement artifacts. Indeed, comparing output signals from all the four systems with the output signal of the spirometer, it is worthwhile pointing out that spirometer is much less sensitive to movements. All the signals are bandpass filtered (0.1–1 Hz), dc filtered, and normalized. We could have used the time-domain processing to extract the respiration frequency, but we preferred to do that in the frequency domain because it provides more precise and reliable results.

Frequency-domain analysis was done to estimate the fast Fourier transform (FFT) of the output signals from all the systems used here. The frequency at which the spectrum exhibits the maximum corresponds to the respiratory frequency, being the main harmonic of the signal. Moreover, we calculated and compared power spectral density (PSD) when subjects do the same controlled physical exercise as AMNB, but this time, arm movement breathing (AMB) and when subject normally breath without moving.

This comparison analysis aims at evaluating the bandwidth associated with movement and respiration activities. Results show that movement frequency falls into the same bandwidth as the breath. Therefore, a simple filtering, even though very selective, is not sufficient to reject the movement artifact. This confirms that, in contrast to spirometer, indirect methods to monitor respiration activity are heavily and intrinsically affected by movement artifacts. The purpose of this study is to identify which one is less prone to these drawbacks.

4) Statistics: When two or more measurement techniques are compared in order to find which has the best performance, a statistical investigation is necessary in order to provide a specific confidence interval (CI) in which the results fall [22]. Concerning the measurement method comparison, previous studies in the literature show several approaches: the use of comparison of mean, correlation coefficient, regression analysis, Bland–Altman (BA) plots, and other [23]–[25]. From the examples reported in the literature [26], [27], the combination of CIs, test of hypotheses, and BA plots can provide exhaustive factors to determine which method is the best. Here, we used comparison of means (statistical hypothesis test and CIs) and BA plots. Data treatment before applying statistical techniques is briefly described. The acquired data, after being resampled and filtered, were divided into 30-s intervals where each interval was overlapped by 30%; hereafter, we will refer to these intervals as tries. The FFT algorithm was launched on each try and its maximum was taken as a breathing frequency component. In this way, the value of breathing frequency is the result of an estimation of all the tries considered for each subject. This technique is an evolution of Periodogramm technique that is performed to reduce the effect of noise measurement as reported by Schuster [28]. The obtained results constitute data for statistical analysis. The first statistical approach is based on performing a t-test on the hypothesis that two independent samples come from distributions with equal means. Before performing the t-test, we validated the assumption of the normal distribution of the data with the Kolmogorov–Smirnov test with the Lilliefors correction. In other words, the t-test assesses whether the means of two groups are statistically different from each other. The data are assumed to come from normal distributions with unknown variance. This test returns the p-value, i.e., the probability that a difference between groups during an experiment happened by chance. The lower the p-value, the more likely it is that the difference between groups was caused by chance. Groups in this paper are represented by four pairs of datasets of breathing frequencies. One dataset of each pair is always represented by respiration frequencies coming from the spirometer, while the other dataset is represented in rotation by the respiration frequencies identified by the other systems. The second statistical approach was to evaluate BA plots, commonly used for assessing agreement between two methods of clinical measurement. They plot the difference between the two measurements on the Y-axis and the average of the two measurements on the X-axis, and are commonly used in method comparison studies. This technique is based on the conjecture that if the measurements are made equivalently, the best estimation is given by the average. A high correlation for any two methods designed to measure the same property does not automatically imply that there is good agreement between the two methods.

III. Experimental Results and Discussion

Results from the time- and frequency-domain analyses and filter application, as well as results from statistical analysis are reported and discussed here. As described in Section II, the time-domain analysis was used to demonstrate the hypothesis that spirometer was not affected by movement artifacts, and therefore can be considered as gold standard. Fig. 6 shows the signals simultaneously acquired from all the systems, during a controlled movement such as AMB, but without breathing (AMNB). If the systems were sensitive only to breathing and not to movement, their outputs would be expected to be null. As it can be seen in Fig. 6, only the spirometer had a nearly flat response during the movement. This ascertains that spirometer is minimally affected by movement artifacts confirming our hypothesis.

Analysis in the frequency domain is used to estimate the respiration frequency, i.e., the frequency component at which the spectrum exhibits a maximum. In addition, we compared spectra of signals acquired when subjects do a controlled physical exercise with their arms (AMB and when subject normally breath

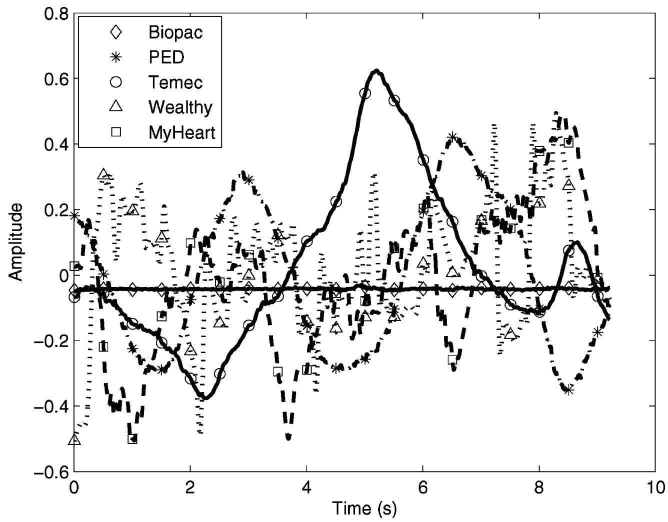

Fig. 6. Output signals over time from all the systems, included the spirometer, during a physical exercise without breathing.

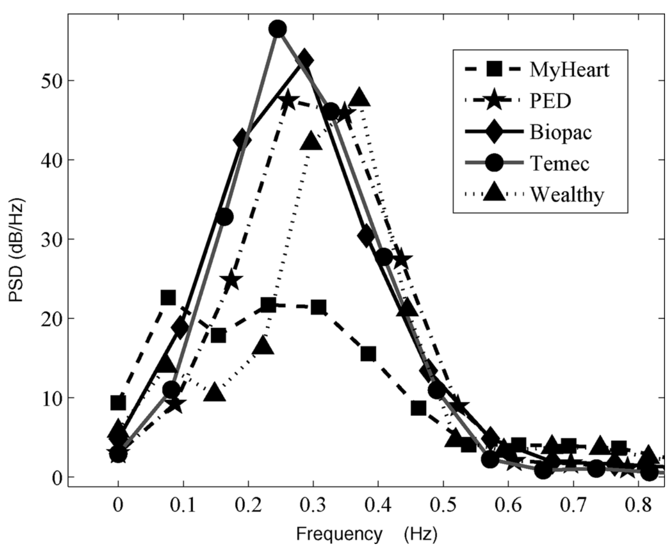

Fig. 7. FFTs of the output signals from all the systems during the AMB.

without moving. Figs. 7 and 8 show FFTs of AMB exercise and a simple respiration activity, respectively, acquired by all the systems over the same time interval. These figures show that movement frequency falls within the same bandwidth as the breath, and a possible filter application, though it could be selective, does not reject movement artifacts. It is worthwhile noting that all the systems exhibit approximately the same frequency peak that reveals the respiratory rate, with small acceptable and expectable variations. Only the FFT curve relative to the MyHeart system during the arm movement shows a little discrepancy. Indeed, the FFT amplitude at the peak frequency associable to the respiration rate is comparable with that of another spurious peak. This ambiguity indicates at a glance a not negligible susceptibility to movement artifacts and we, accordingly, expect that results will show a low capability of the MyHeart system of identifying the respiratory rate when compared with the spirometer, mainly during the physical exercises.

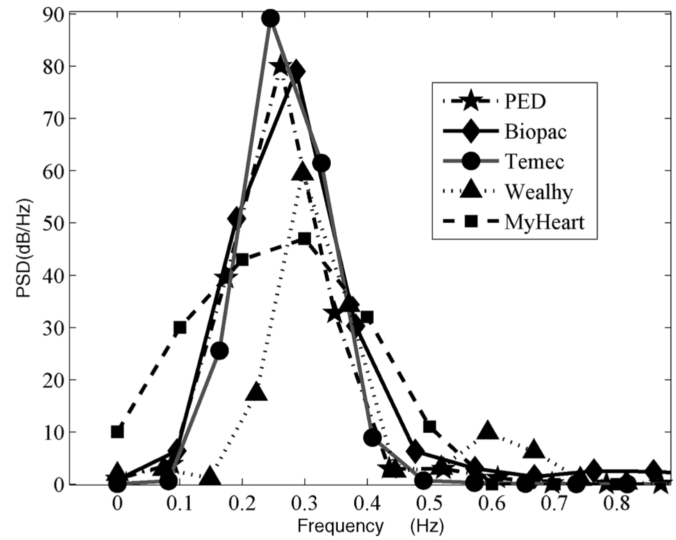

Fig. 8. FFTs of a simple respiration activity acquired by all the systems over the same time interval.

TABLE II
γ-VALUES FOR EACH SYSTEM VERSUS SPIROMETER AND PHYSICAL EXERCISE

Device	Cycle	Elliptic	Running	Walking
Biopac vs PED	0.966	0.7745	0.6643	0.6714
Biopac vs MyHeart	0.3183	0.5618	0.3402	0.4442
Biopac vs Wealthy	0.6431	0.3571	0.4036	0.7909
Biopac vs Temec	0.5599	0.5316	0.4794	0.5476

Concerning statistical analysis, results of t-test are reported in Table II, which show $(1-\alpha)$ values, which we call hereafter γ-values, where α is the significance level, i.e., the probability of rejecting the null hypothesis when it should have been accepted. These values represent the probability of accepting the null hypothesis being true, and were calculated for each system with respect to the spirometer and physical exercises. γ-values indicate the probability of a system to verify the null hypothesis, i.e., how reliably the system is able to identify the breathing frequency.

As it can be noted in Table II, the highest γ-values depend on the exercise we are considering. At first glance, it results highest for *PED* during cycling, walking, and elliptic exercises, while it is highest for *Wealthy* during walking exercise. This would seem to show *PED* as the device that has the highest probability to verify the null hypothesis for most of the physical exercises. Actually, we can surely state there is no evidence against the null hypothesis and data appear to be consistent with it. The γ-values, indeed, are higher than 50% for most of the systems in every exercise. γ-values are, on the other hand, a synthetic parameter averaged on all the signals of each monitoring system and all the subjects. In order to better understand the distribution of data, at least referred to each subject, BA plots are reported.

Figs. 9–12 show BA plots, one for each exercise, where difference between the methods against their means is plotted. On the X-axis of each plot is reported the mean between the spirometer and each system, while on the Y-axis the difference. Into each BA plot, all systems are presented with different markers and only the smallest limits of agreement interval are shown.

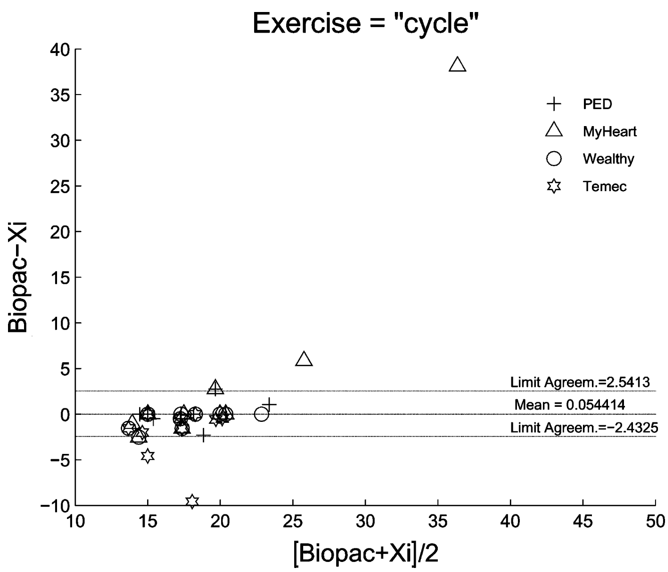

Fig. 9. BA plot relative to the “cycle” exercise, X_i represents in rotation all the wearable systems, PED, MyHeart, Wealthy, and Temec. Each marker in the plot is associated to a subject performing the exercise.

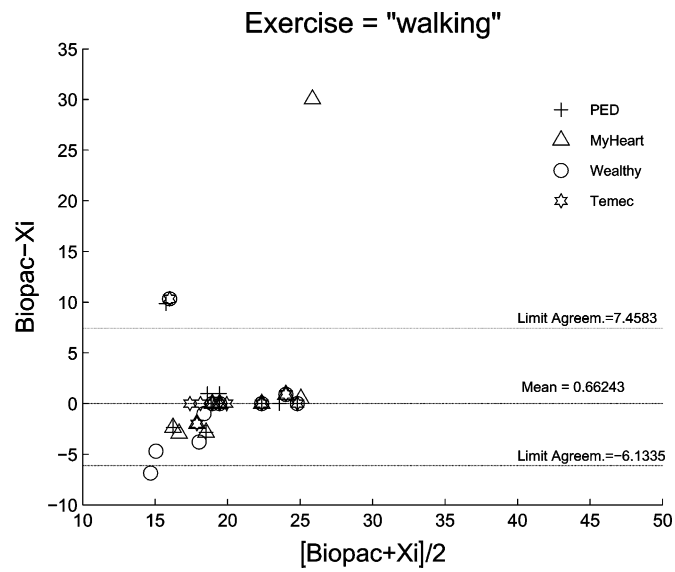

Fig. 11. BA plot relative to the “walking” exercise, X_i represents in rotation all the wearable systems, PED, MyHeart, Whealhy, and Temec. Each marker in the plot is associated to a subject performing the exercise.

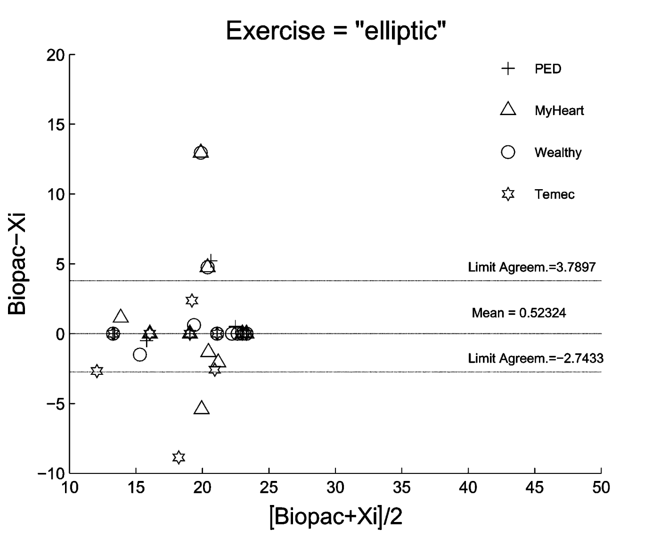

Fig. 10. BA plot relative to the “elliptic” exercise, X_i represents in rotation all the wearable systems, PED, MyHeart, Wealthy, and Temec. Each marker in the plot is associated to a subject performing the exercise.

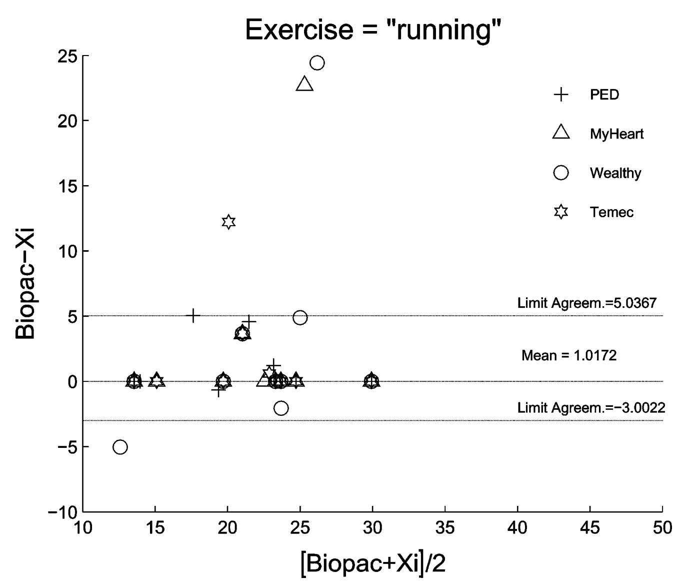

Fig. 12. BA plot relative to the “running” exercise, X_i represents in rotation all the wearable systems, PED, MyHeart, Whealhy, and Temec. Each marker in the plot is associated to a subject performing the exercise.

For limits of agreements, it is intended an interval ranging from $-2s$ to $+2s$ with respect to the mean value, where s is the standard deviation. Those markers that are outside the limits of agreement are denoted as outliers and indicate lack of agreement between the particular indirect measurement system and the spirometer. It is worthwhile noting that *PED* remains within these limits of agreement for almost all exercises. In addition, the plot of differences against means also allows us to investigate any possible relationship between the measurement error and the true value. For this study, a normal data distribution is supposed to exist. Analyzing the plots more in details, we can note that relative to cycle exercise, we have a few of outliers associated to MyHeart and Temec systems. In the elliptic exercise, there are more outliers, exhibited by MyHeart and Temec, as well as Wealthy systems. In the walking exercise, most of the Wealthy systems are contained within the limits of agreement. Finally, in the running exercise, the band of agreement contains most the data, except for MyHeart, Temec, and Wealthy. These differences can be justified by the fact that each exercise can induce some specific movement that can be easily detected by a monitoring system rather another one.

In order to identify with a reasonable statistical significance which indirect monitoring system is most similar to the spirometer, we have to also provide the CIs centered on the mean values. In Table III, CIs for each indirect monitoring system versus the spirometer and with respect to the physical exercise are reported.

TABLE III
CIs OF EACH SYSTEM VERSUS SPIROMETER AND PHYSICAL EXERCISE

Device	Cycle	Elliptic	Running	Walking
Biopac vs PED	0.077	0.100	0.124	0.219
Biopac vs MyHeart	0.535	0.264	0.491	0.599
Biopac vs Wealthy	0.156	0.284	0.347	0.204
Biopac vs Temec	0.279	0.255	0.320	0.430

It is worthwhile pointing out that *PED* has the narrowest interval of agreement for cycle, elliptic, and running exercises while it becomes wider than the *Wealthy* during walking exercise only. These results agree with the γ-values.

IV. CONCLUSION

In this study, we compared the performance of different respiratory monitoring systems. From a visual inspection of the normalized signal and the FFT transformed signal, we can see that, in basal condition, each of the instruments has a similar response, but when subjects are moving, sensitivity to the motion artifacts increases, especially if movements are at the thorax level; the piezoelectric belt system is more robust to such a kind of artifacts. This fact is validated by results of the statistical analysis. Detailed investigation on the reasons of this observed fact will be done in the future.

In BA plots, the differences among data, which are plotted against their means, follow a normal distribution because a lot of the variations among subjects are removed and only the measurement errors are left. The measurements themselves do not have to follow a normal distribution, and often they do not. Normal distribution of the differences can be checked by plotting a histogram. If this is skewed or has very long tails, the assumption of normality may not be valid.

On the basis of the obtained results, we can draw some conclusions. First, the PED shows the best characteristics to be used for breathing frequency monitoring, even when the subject is moving. All the movements in all the exercise are controlled and kept as constant as possible, and this introduces only a few frequency components in the same bandwidth as the breath. PED uses as piezoelectric sensor a cable that is wrapped around the thorax. This specific intrinsic configuration could dramatically reduce noise effect due to movement. Future plans aim at embed this sensor into a shirt. Some preliminary attempts were done within the Proetex project (FP6-2004-IST-4-026987).

The Wealthy system as well as the MyHeart shirt were conceived to fight cardiovascular diseases, and they have been optimized for the cardiac monitoring. In particular, the Wealthy shirt aims to monitor cardiac patients during rehabilitation phase after a surgical intervention, and as a matter of fact, the best results over the breathing experiments have been detected during the walking exercises. It should be pointed out that none of the systems tested in this paper have been developed with motion artifact rejection as a primary testing specification. More attention has been devoted to issues such as wearability, comfort, and low cost. So, ample margins of improvements in performance in terms of susceptibility to motion artifacts can be expected.

A key factor in achieving success will be signal processing and development of more complex algorithms than those presented here is required. This will be the subject of future work together with the refinement of sensing techniques and body placement.

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