Design and Implementation of Sensing Shirt for Ambulatory Cardiopulmonary Monitoring
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Abstract
Ambulatory monitoring of cardiopulmonary parameters during wake, sleep, and activity states can expand our understanding of health and diseases. In this paper, we present a wearable cardiopulmonary monitoring system called the ‘Sensing Shirt’. The Sensing Shirt consists of a T-shirt with sensors integrated for physiological monitoring, a data acquisition unit (DAQU), and a set of PC-based software. A number of vital signs such as electrocardiogram (ECG), rib cage (RC) and abdominal (AB) respiration, photoplethysmogram (PPG), SpO₂, and posture/activities can be acquired by the shirt. The physiological data are stored in a microSD memory card, and then analyzed offline to extract parameters such as heart rate (HR), breathing rate, SpO₂, tidal volume (TV), pulse wave transit time (PWTT), and respiratory sinus arrhythmia (RSA). In the Sensing Shirt system, the ECG is acquired by means of active electrodes. PPG and ECG are sampled by one high-speed 16-bit, analog-to-digital converter at 500 Hz for precise PWTT calculation. Technique of pulse amplitude modulation is used for signal conditioning both in respiratory inductive plethysmography and PPG circuits to reduce the power consumption. Powered by 2 AA batteries with 1600 mAh capacity, the whole system can work more than 48 h continuously without interruption. Basic performance tests demonstrate that prototypes of this system can capture ECG RC and AB respiration, and PPG signals effectively, and extract parameters such as HR, breathing rate, SpO₂, and TV accurately. This advanced ambulatory monitoring system may be used in both home healthcare and scientific research.

Keywords: Wearable, Cardiopulmonary monitoring, Pulse amplitude modulation, Active electrode, Respiratory inductive plethysmography, Pulse wave transit time

1. Introduction
Continuous monitoring of physiological and physical parameters is necessary for the assessment and management of personal health status [1]. It contributes significantly to the reduction of healthcare cost and ensuring those who need urgent care get it sooner. In recent years, there has been an increasing interest in wearable physiological monitoring systems, both in research and commercial areas. Many novel designs for wearable sensors and systems have been developed, such as AMON [2], SenseWear [3], LifeShirt [4], Smart Shirt [5], and Smart Vest [6]. Wearable health monitoring systems employ wearable sensors to acquire vital signs from human bodies such as heart rate (HR), respiration, body temperature, and blood pressure. Wrist, belt and smart clothes are frequently used for placement of biomedical sensors and devices for ambulatory use. The field of smart textile-based wearable biomedical systems has been generating a lot of interest in the research and business communities [7,8], with the aims of getting the technology into the marketplace.

Cardiopulmonary function monitoring is an essential part for a health-monitoring system. Chronic diseases such as chronic heart failure, atrial fibrillation, pulmonary emphysema, and chronic obstructive pulmonary disease (COPD) represent a considerable workload in primary care. There is also interest as to whether it would be feasible to conduct sleep disorder testing at home. LifeShirt is a currently commercially available monitoring system capable of cardiopulmonary physiological data capture [9,10]. The key technology of this system is its patented respiratory inductive plethysmography (RIP). A detailed quantification of volume, timing and shape parameters in the respiratory pattern waveform can be derived from RIP technology. LifeShirt opened a new era for comprehensive monitoring of cardiopulmonary physiology, but this system...
acquires electrocardiogram (ECG) with the current passive Ag-AgCl electrodes, which has the disadvantages of baseline wander, power line interference, and movement artifact noise during ambulatory use [11]. As the pulse oximeter is integrated in the system by the serial port, it is hard to synchronize the photoplethysmogram (PPG) data received from the series port with the LifeShirt's intrinsic ECG signal to derive an accurate pulse wave transit time (PWTT), which is a useful parameter both for sleep apnea detection and continuous blood pressure estimation. Furthermore, as Life Shirt is a PDA-based data acquisition system, it is not convenient for users to carry this device in ambulatory condition. Battery life and cost of this system are also problems.

In this paper, we present a microcontroller-based wearable, cardiopulmonary monitoring system called Sensing Shirt, which incorporates our research on wearable technology such as active electrodes for ambulatory ECG sensing, technique of pulse amplitude modulation (PAM) for RIP [12], and low-power PPG. The whole system consists of a T-shirt with sensors integrated, a data acquisition unit (DAQU), and PC-based software for data processing and display. The system integrates sensors and cables into an elastic, lightweight shirt, and sensors are distributed on several specific body locations to accurately measure the vital parameters. The DAQU is placed in a little pocket on the outer surface of the shirt, and a removable microSD card is incorporated in the DAQU for data recording. Vital parameters such as single-lead ECG, rib cage (RC) and abdominal (AB) respiration, tidal volume (TV), PPG, HR, SpO2, PWTT, respiratory sinus arrhythmia (RSA), and posture/activities can be acquired by this system.

In the Sensing Shirt system, the PPG signal is sampled along with other cardiopulmonary parameters by one high-speed, 16-bit ADC. Synchronized AD conversion between PPG and ECG signals contributes to more accurate PWTT calculation. As the power consumption is a crucial factor for a wearable system powered by batteries, the technique of PAM is used both for RIP and PPG signal conditioning, with the microcontroller generating control signals, to reduce the power consumption dramatically in the analog section. Powered by two AA alkaline batteries with 1600 mAh capacity, the Sensing Shirt system can work continuously more than 48 h without interruption. Basic performance tests show that prototypes of this system can capture ECG, RC and AB respiration, and PPG signals effectively, and extract parameters such as HR, breathing rate, SpO2, and TV accurately.

2. Materials and methods

The Sensing Shirt system consists of three parts: a comfortable T-shirt with embedded sensors to capture the physiological signals, a DAQU for data acquisition and SpO2 calculation, and a set of PC-based software for offline parameter extraction. All data sampled is stored on a microSD memory card and transferred to PC with a USB card reader. Data analysis and display are performed on a PC with software developed on the LabVIEW 8.2 software platform (National Instruments, Austin, TX).

2.1 Sensing shirt

Basic cardiopulmonary parameters such as ECG, RC and AB respiration, PPG, and SpO2 can be acquired by the shirt. Figure 1 illustrates the Sensing Shirt with sensors integrated for collecting multiple cardiopulmonary parameters. A 3-axis accelerometer is embedded in the DAQU to record body posture and activities. ECG button electrodes and RIP inductive wires are integrated in the fabric of the Sensing Shirt and connected to the DAQU by wires, which are also stitched into the fabric. These sensors are placed at specific body locations and integrated into the shirt in a manner which makes them stay in good contact with the user’s body. RIP sensors are woven into the shirt around the user’s chest and abdomen. Two parallel sinusoidal arrays of insulated wires embedded in elastic bands are woven into the shirt. For each RIP sensing band, tightening devices of Velcro tape for individual adjustment of tightness of the band is equipped to prevent its longitudinal movement over the surface of the torso of body. Considering the convenience of handling and comfort of wearing, the ECG button electrodes are embedded at special locations (RA, RL, LL, as shown in Fig. 1) in the inner surface of the shirt, and wires are stitched into the shirt. Comfortable materials blending cotton and knitted lycra were chosen to fabricate the shirt. Wearable sensors integrated in the shirt are attached via one cable tree and a secure connector to the DAQU. That makes it easy for the ordinary user to plug and play, and avoid operation errors. PPG sensor placed at the index finger/ear lobe is plugged into the DAQU through another different cable and secure connector.

2.2 Data acquisition unit

The DAQU is light-weight so as to be placed into a little pocket on the outer surface of the shirt, as shown in Fig. 1. Figure 2 illustrates the block diagram of DAQU, which includes the complete electronic components for signal conditioning, digitization, processing, and recording. The DAQU is designed to keep its size and power consumption to a minimum. A commercial microSD memory card with the capacity of 1G was chosen as the storage medium for its tiny dimension. An ARM Cortex-M0 based, low-power 32-bit
MCU LPC1114 (NXP Semiconductors N.V., Eindhoven, The Netherlands) performs the AD conversion, control signal generation for PAM used in RIP and PPG analogy circuit, signal processing for SpO₂ calculation, and data storage in the microSD card. An embedded real-time operating system (RTOS) UCOS-II (version 2.86) runs on the ARM core for multiple tasks scheduling. Multiple analog signals are sampled by a 16-bit ADC ADS8320 (Texas Instruments, MN, USA) through the synchronous serial interface with the MCU. Eight-channel analog multiplexers ADG658 (Analog Devices, Inc., Mass., USA) act as channel selection controlled by the MCU. The system is powered by two AA batteries (LR6). The voltage of the battery is raised to 3.4 V by a step-up DC-DC converter MAX1722, and two LDO chips CAT6217-3.3 are used to generate two separate 3.3-V DC supplies feeding to analog and digital circuits, respectively.

2.3 Sensor unit

All analog signal conditioning circuits are on one PCB board placed in the DAQU. Analog signals from the sensors are conditioned in the DAQU to levels suitable for digitization and processing. Though some special ECG sensors, such as silicon rubber electrodes with pure silver fillings, fabric electrodes, filled gold electrodes, have been developed [13], these kinds of electrodes are prone to contact impedance variation, baseline wander, and movement artifact noise when the user is in motion. In the Sensing Shirt system, conventional Ag-AgCl electrodes are used for ECG sensing for their high reliability. In order to improve the ECG signal quality for ambulatory use, active electrodes are incorporated in the Sensing Shirt. An active electrode with its biopotential amplifier embedded in the ECG button electrodes, as shown in Fig. 1, performs the signal amplification as close to the transducer as possible, contributing to a high-quality ECG signal by its ultra-high input impedance and low output impedance.

In our system, the gain of the amplifier circuit of the active electrode is set to one, and it actually acts as an emitter follower. The system can acquire one-lead ECG (standard lead-II) using three Ag-AgCl electrodes. On the analogy circuit board in the DAQU, a differential preamplifier with a gain of 8, a high-pass filter with 0.5 Hz cutoff frequency, a fourth-order Butterworth low-pass filter with 100-Hz cutoff frequency, and a main amplifier with a gain of 125 are used to acquire clear ECG signal. As the shirt is a battery-powered system and active electrodes have been used, 50-Hz notch filter in hardware is omitted. Digital signal processing technology is used in the PC-based software for digital filtering, scaling, etc. The ECG signal is sampled at 500 Hz. The HR is measured from the ECG signal, and R-waves are detected with 2-millisecond resolution.

RIP sensors are integrated in the Sensing Shirt for respiration monitoring. Comparing to impedance pneumography, inductive plethysmography has more advantages, including greater accuracy, better sensitivity and better security for patients. The Sensing Shirt system incorporates our patented RIP with PAM technique to detect respiration with high signal quality and ultra-low power consumption [12]. RIP sensors are excited by one high constant-current source momentarily and periodically, and in a very short time, the respiration signals are detected and sampled. The RC and AB sensors are excited consecutively under the control of sequence signal to avoid crosstalk between each other. A low-voltage and low-on-resistance CMOS analog switch is employed for channel selection controlled by a digital gating signal applied to its control gate. In order to improve the integration and reduce the system power consumption, control signals for PAM are generated by the MCU through its general-purpose input and output ports.
The HR is calculated by analyzing the ECG waveform to determine the QRS complex beat by beat. Studies on automatic detection of QRS complex have been carried out for many years [16]. We detect the QRS complex by the first-derivative and threshold-comparison approach. Firstly, the power-line interference is removed by a subtraction procedure [17], if necessary. The procedure is simple and has good performance in coping with changes in amplitude of the power-line interference. Then the QRS complex is detected by a band-pass filter followed by a threshold comparison algorithm. Finally the R-wave position is located and saved, and the RR interval is calculated. As the sampling rate of the ECG signal is 500 Hz, it is easy to calculate the HR as follows:

$$HR = 60/(RR \text{ interval} / 500).$$

When correctly calibrated, RIP allows the measurements of volume and time components of breathing cycles as well as the relative contribution of RC and AB. Some calibration techniques, based on the two-compartment model of the respiratory system model by Konno and Mead [18], have been developed, as shown in the following relationship:

$$\Delta Va = \Delta Vrc + \Delta Vab$$

(1)

The change of volume measured at the mouth (\(\Delta Va\)) is the sum of an RC (\(\Delta Vrc\)) and an AB (\(\Delta Vab\)) contribution. These volume changes are obtained indirectly by measuring the variations of two representative thoracic and abdominal dimensions by RIP. Equation 1 can be rewritten as:

$$\Delta Va = \alpha \Delta RC + \beta \Delta AB$$

(2)

where \(\Delta RC\) and \(\Delta AB\) are changes (\(\Delta\)) in RIP ribcage and abdomen signals. \(\alpha\) and \(\beta\) are calibration coefficients. Typically, RIP calibration involves collecting data to ascertain \(\alpha\) and \(\beta\). The \(\Delta Va\), RC excursion (\(\Delta RC\)), and AB excursion (\(\Delta AB\)) are simultaneously recorded. These values are then used to set RC and AB channel amplifier gains (coefficients \(\alpha\) and \(\beta\)) for implementing equation (2) during ventilation monitoring. After calibration procedure, the breath-by-breath TV can be estimated according to equation (2) by RC and \(\Delta AB\).

Least squares method (LSQ) calibration [19] and qualitative diagnostic calibration (QDC) procedure [20] are commonly used for RIP calibration. We utilize the LSQ method for more accurate ambulatory TV estimations and the QDC method for relative TV estimations during sleeping periods. QDC calibration procedure does not require separate calibrations for upright and supine position of the torso, and can be finished during a 5-min quiet period of breathing. This calibration method is preferred during sleeping periods.

The calculation of \(\text{SpO}_2\) is performed in real time in the MCU, and the \(\text{SpO}_2\) values are stored with other physiological parameters simultaneously. The approach of first-derivative and threshold-comparison is used to process each PPG signal to determine the occurrence of every pulse [21]. Then the peak-to-peak value of each PPG signal, pulse by pulse, is calculated by the peak detection method of finding the local maxima and minima of each pulse. After calibration, the normalized ratio between two wavelengths is used to predict \(\text{SpO}_2\) values by checking the conversion table, which converts the ratio of normalized absorbance into percent \(\text{SpO}_2\).
2.5 Performance test

A performance test was completed to study the consistency and reliability of the physiological data acquired by the Sensing Shirt. For ECG signal, the effectiveness of QRS detection algorithm was firstly evaluated. ECG signals from the MIT/BIH arrhythmia database were chosen as the testing input to the off-line analysis software for evaluation. We extracted ECG segment of 5 minutes’ duration randomly from each file in the database for test. Then we used a patient simulator (Metrology Services, DNI Nevada ECG 200, USA) to generate typical 1-mV peak-to-peak normal ECG signals to the DAQRU through three Ag-AgCl electrodes. HR was changed manually from 30 to 240 BPM (beat per minute). Signals were recorded in the microSD memory card and an off-line analysis was performed to test the reliability and accuracy.

Because the RIP measures the changes of cross-section area, there is no standardized apparatus to test its performance in clinical environment. We developed a relatively standard method for Sensing Shirt’s RIP testing. A medical ventilator (Siemens, Servo-i, Germany) and a lung simulator (LS-2000®, BC Biomedical) were used to make the changes of cross-section area both in RC and AB positions. The ventilator mechanically pumped air into and out of the lung simulator, which was placed under the shirt. Constant 300-mL TV was generated by the ventilator periodically, so the volume of lung simulator changed under the control of the ventilator to generate a reproducible signal to the RIP sensors. Different breathing rates were set manually through the operation menu on the ventilator’s screen. Data were recorded in the microSD memory card for an off-line breathing rate calculation.

We used a SpO2 analyzer (Fluke Biomedical, daeg, Norway) to verify the accuracy of SpO2 measurement. A conversion table which converts the ratio of normalized absorbance into percent SpO2 was obtained from the simulator.

After the preliminary check by the patient simulator, we carried out two human body experiments for ECG and RIP evaluation, respectively, with the standard measurement methods as reference. The accuracy of TV estimation by the Sensing Shirt was firstly evaluated. The experiment was conducted on 15 healthy male subjects aged from 20 to 40 from whom informed consent had been obtained. The validation of the device was carried out for about 30 min on each subject in supine position. RC and AB excursion and respiratory flow were simultaneously recorded by the Sensing Shirt and the ventilator tester (imtmedical, FlowAnalyser™ PF-300, Switzerland) respectively, with the subject wearing a face mask sealed around the nose and mouth. QDC calibration procedure was firstly used to determine the calibration coefficients for TV measurement, and then breath-by-breath TV was compared between the two systems. The absolute percentage error, correlation coefficient, and linear regression coefficient on breath-by-breath TV between the two systems were calculated to estimate the performance of the Sensing Shirt.

We carried out treadmill exercise and cycle ergometer exercise test for dynamic ECG evaluation, with the American Thoracic Society/American College of Chest Physicians recommendation for reference [22]. Ten healthy men (no known respiratory or cardiac disease) aged from 22 to 32 participated in this test, having given prior consent. The raw ECG data were recorded by the Sensing Shirt, and the beat-by-beat HR was stored simultaneously by an HR monitor (Polar Sport Tester, RS800, Finland) during two sequential conditions. Firstly, the participants completed an incremental treadmill test, which started in stationary standing state (0 mph) and continued in walking state (4 mph), jogging state (6 mph), and fast running state (8 mph). Each state lasted for 2 minutes. Secondly, they completed an incremental cycle ergometer exercise, which started in stationary sitting state on the stationary bicycle (0 rpm) and continued in slow pedaling state (20 rpm), medial pedaling state (30 rpm), and fast pedaling state (40 rpm). Each state lasted for 2 minutes. In the section of data processing, for each exercise state, the raw ECG data captured by the Sensing Shirt was firstly processed to derive HR, and then the mean HR of the two systems was calculated. The mean HR was analyzed statistically according to the method described by Bland–Altman [23] and plotted as the difference in measurements against the mean of the two measurements. The average of the differences between the two methods was used to calculate the overall bias and the standard deviation (SD) of the differences between the methods was used to calculate the precision. The bias and precision were analyzed to estimate the performance of the device.

3. Results

The Sensing Shirt has been developed and tested for functionality and comfort levels for users. The DAQU weighs 104 g, with the three dimensions of 102 mm × 64 mm × 20 mm. Figure 3 illustrates the physiological data captured by Sensing Shirt during an incremental treadmill exercise. Figure 4 illustrates the screen shot of the off-line data analysis software with waveform of ECG, RC, AB, PPG, SpO2 and motions.
Tables 1 and 2 show the test results of heat rate and respiratory rate evaluation, respectively. Table 3 shows the test results for the QRS detection algorithm, which produces 6 false-positive detections (FP), 4 false-negative detections (FN), and 7303 true-positive detections (TP) in the total 7407 QRS complexes (QT). This represented sensitivity (Se) of 99.95% and positive predictive accuracy (Ppa) of 99.92%.

When SpO2 was predicted with the simulator probe in the range between 35% and 100%, the correlation coefficient was 0.999 and the SD of percent SpO2 was 0.15. The maximal absolute difference between the SpO2 value and the simulator during the whole test was 1%.

For TV measurement comparison, sequential TV collected from the Sensing Shirt and PF-300 were aligned in adjacent columns of a data spreadsheet for each participant for evaluation. Of totally 11,437 validation breaths, 93.85% (10,734 breaths) of the breath-by-breath TV measured by

Table 1. Results of heart rate test.

<table>
<thead>
<tr>
<th>Patient simulator (ECG 200)</th>
<th>30 BPM</th>
<th>60 BPM</th>
<th>120 BPM</th>
<th>180 BPM</th>
<th>240 BPM</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sensing shirt</td>
<td>30.0 ± 0</td>
<td>60.0 ± 0</td>
<td>120.0 ± 0</td>
<td>179.1 ± 0.3</td>
<td>240.0 ± 0.2</td>
</tr>
</tbody>
</table>

Table 2. Results of breathing rate test.

<table>
<thead>
<tr>
<th>Ventilator (SERVO-1)</th>
<th>4 BPM</th>
<th>6 BPM</th>
<th>10 BPM</th>
<th>15 BPM</th>
<th>20 BPM</th>
</tr>
</thead>
<tbody>
<tr>
<td>Sensing shirt</td>
<td>3.9 ± 0.1</td>
<td>5.9 ± 0.2</td>
<td>10.0 ± 0.1</td>
<td>15.0 ± 0.0</td>
<td>20.0 ± 0.2</td>
</tr>
<tr>
<td>Ventilator (SERVO-2)</td>
<td>25 BPM</td>
<td>30 BPM</td>
<td>40 BPM</td>
<td>50 BPM</td>
<td>60 BPM</td>
</tr>
<tr>
<td>Sensing shirt</td>
<td>25.0 ± 0.2</td>
<td>30.0 ± 0.7</td>
<td>39.9 ± 0.5</td>
<td>51.2 ± 3.8</td>
<td>59.8 ± 5.1</td>
</tr>
<tr>
<td>Ventilator (SERVO-3)</td>
<td>70 BPM</td>
<td>80 BPM</td>
<td>90 BPM</td>
<td>100 BPM</td>
<td></td>
</tr>
<tr>
<td>Sensing shirt</td>
<td>69.5 ± 3.0</td>
<td>79.7 ± 2.6</td>
<td>89.5 ± 2.1</td>
<td>100.5 ± 4.3</td>
<td></td>
</tr>
</tbody>
</table>

Sensing Shirt were shown to be within ±10% of simultaneous PF-300 measurements, 98.10% (11,220 breaths) within 15%, and 99.03% (11326 breaths) within ±20%. The result was comparable with others’ report [24]. Table 4 shows the test results of breath-by-breath TV comparison between two systems.

The Bland–Altman plot for the HR detection during incremental exercises is illustrated in Fig. 5. The bias for HR detection during exercises was 0.24, and precision was 1.54. There existed systematic difference on the mean HR
Some of the extracted parameters such as HR, breathing rate, TV, and SpO2 have been validated by the preliminary check, and the reliability of HR detection during exercises was also tested. Figure 3 shows the physiological data captured by the Sensing Shirt at different movement stages during an incremental treadmill exercise. Though more movement artifact noise was imposed on the physiological data with the increment of exercise intensity (which can be affirmed by the signals of X, Y, and Z from the 3-axis accelerometer), the Sensing Shirt system could capture the physiological data and extract basic parameters effectively. For HR detection, though statistically, there is systematic difference in mean HR between the Sensing Shirt system and the Polar Sport Tester RS800, the mean difference is very small (0.24). This result is comparable with others’ report with respect to the validity and reliability of cardiorespiratory measurements recorded by the Life Shirt during exercise [25]. A comparison of HR detection between the Life Shirt and Polar Sport Tester also demonstrated there was significant difference between systems, and the mean difference was also small. The Bland–Altman plot in Fig. 5 shows all the HR were within the mean ± 2SD range, which signifies good stability in HR detection during exercises.

4. Discussion

This work contributes to the state of the art for wearable, ambulatory healthcare monitoring, by integration of multiple sensors into the shirt. The Sensing Shirt system is a logical extension of ECG monitors, with more physiological parameters, more patient comfort and more ease of use. Our wearable sensor unit has been carefully designed to decrease the power consumption and improve the signal quality for ambulatory use. Sensing Shirt represents our initial efforts in developing wearable technology for non-intrusive cardiopulmonary monitoring. Performance tests on these basic parameters demonstrated the accuracy and reliability of this prototype system.

Table 4. Results of breath-by-breath TV comparison between two systems.

<table>
<thead>
<tr>
<th>Participant</th>
<th>No. of breaths</th>
<th>Error &gt; ±10%</th>
<th>Error &gt; ±15%</th>
<th>Error &gt; ±20%</th>
<th>Correlation coefficient</th>
<th>Regression coefficient</th>
</tr>
</thead>
<tbody>
<tr>
<td>1</td>
<td>700</td>
<td>4.78%</td>
<td>0.72%</td>
<td>0.29%</td>
<td>0.94</td>
<td>1.01</td>
</tr>
<tr>
<td>2</td>
<td>647</td>
<td>5.87%</td>
<td>2.47%</td>
<td>1.70%</td>
<td>0.87</td>
<td>1.00</td>
</tr>
<tr>
<td>3</td>
<td>811</td>
<td>10.73%</td>
<td>3.33%</td>
<td>1.23%</td>
<td>0.96</td>
<td>1.01</td>
</tr>
<tr>
<td>4</td>
<td>864</td>
<td>1.62%</td>
<td>0.46%</td>
<td>0.35%</td>
<td>0.88</td>
<td>1.00</td>
</tr>
<tr>
<td>5</td>
<td>921</td>
<td>3.48%</td>
<td>1.20%</td>
<td>0.76%</td>
<td>0.94</td>
<td>1.01</td>
</tr>
<tr>
<td>6</td>
<td>858</td>
<td>9.91%</td>
<td>1.75%</td>
<td>0.35%</td>
<td>0.85</td>
<td>1.00</td>
</tr>
<tr>
<td>7</td>
<td>741</td>
<td>10.95%</td>
<td>3.92%</td>
<td>1.89%</td>
<td>0.90</td>
<td>1.01</td>
</tr>
<tr>
<td>8</td>
<td>1073</td>
<td>5.23%</td>
<td>1.59%</td>
<td>0.84%</td>
<td>0.84</td>
<td>1.00</td>
</tr>
<tr>
<td>9</td>
<td>742</td>
<td>3.10%</td>
<td>0.54%</td>
<td>0.27%</td>
<td>0.85</td>
<td>1.02</td>
</tr>
<tr>
<td>10</td>
<td>915</td>
<td>7.76%</td>
<td>2.51%</td>
<td>1.64%</td>
<td>0.87</td>
<td>1.01</td>
</tr>
<tr>
<td>11</td>
<td>584</td>
<td>10.27%</td>
<td>3.25%</td>
<td>1.02%</td>
<td>0.88</td>
<td>0.98</td>
</tr>
<tr>
<td>12</td>
<td>519</td>
<td>5.97%</td>
<td>3.08%</td>
<td>2.31%</td>
<td>0.85</td>
<td>1.00</td>
</tr>
<tr>
<td>13</td>
<td>692</td>
<td>4.64%</td>
<td>2.31%</td>
<td>1.30%</td>
<td>0.96</td>
<td>0.98</td>
</tr>
<tr>
<td>14</td>
<td>952</td>
<td>1.94%</td>
<td>0.21%</td>
<td>0.11%</td>
<td>0.91</td>
<td>1.00</td>
</tr>
<tr>
<td>15</td>
<td>418</td>
<td>9.70%</td>
<td>3.15%</td>
<td>1.69%</td>
<td>0.98</td>
<td>1.00</td>
</tr>
<tr>
<td>Total</td>
<td>11437</td>
<td>6.15%</td>
<td>1.90%</td>
<td>0.97%</td>
<td>0.90</td>
<td>1.00</td>
</tr>
</tbody>
</table>

Figure 5. Bland-Altman plots for HR comparison from 10 subjects during incremental treadmill and cycle ergometer exercises. The limits of agreement defined by 2SD above and below the bias are shown by dashed lines.
these two parameters [27,28]. In clinical practice, the use of PSG recording is a standard evaluation procedure for sleep-related breathing disorders. However, PSG is not suitable for long-term monitoring in the home environment. As electroencephalogram (EEG) is not integrated in most ambulatory monitoring system, PWTT will improve the detection of sleep respiratory events and microarousals during sleep testing at home [29]. A number of rigidly controlled experiments will be carried out between the PSG and the Sensing Shirt to evaluate their sensitivity and specificity for sleep-event detection.

The Sensing Shirt is able to simultaneously monitor several physiological systems, thus allowing for assessments of the interactions among various physiological parameters. This system offers us a useful tool to acquire circadian cardiopulmonary physiological parameters. The cardiorespiratory interaction and its changes during the diurnal cycle will provide detailed information on the autonomic nerve system status [30]. Data mining from these long-time physiological parameters to give explicit explanation about health status is another challenge for us.

In the Sensing Shirt’s current implementation, raw data is off-line processed on a PC platform but not in real time. This may limit the possibilities for detailed inspection of the data by the researcher and physicians. In the next step, a Bluetooth- and Smartphone-based telenotifying system will be developed. With the Bluetooth module integrated in the Sensing Shirt, the telenotifying application will work on any mobile phone that can run Java, and data can be transmitted either over a wireless or cellular network. As the Sensing Shirt prototypes have demonstrated the feasibility of the concept and solutions to the key technologies utilized in wearable sensors and systems, mobile phone-based tele-monitoring system will be relatively easy to be developed.

5. Conclusions

A wearable cardiopulmonary monitoring system called Sensing Shirt has been developed with wearable sensors integrated in an elastic T-shirt. A number of vital signs, including ECG, RC and AB respiration, PPG, SP02, and posture/activities, can be acquired by this system. The technique of PAM used in the RIP and PPG sections can reduce the power consumption effectively, and synchronized AD conversion between ECG and PPG contributes to precise PWTT calculation. Powered by 2 AA batteries with 1600-mAh capacity, the whole system can work more than 48 h continuously without interruption. Performance tests on these basic parameters demonstrated the accuracy and reliability of this prototype system. This advanced ambulatory monitoring system may be used in both home healthcare and scientific research.

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References

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