DESIGN AND IMPLEMENTATION OF A WEARABLE, MULTIPARAMETER PHYSIOLOGICAL MONITORING SYSTEM FOR THE STUDY OF HUMAN HEAT STRESS, COLD STRESS, AND THERMAL COMFORT

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To cite this article: Yu-Hong Shen, Jie-Wen Zheng, Zheng-Bo Zhang & Chen-Ming Li (2012) DESIGN AND IMPLEMENTATION OF A WEARABLE, MULTIPARAMETER PHYSIOLOGICAL MONITORING SYSTEM FOR THE STUDY OF HUMAN HEAT STRESS, COLD STRESS, AND THERMAL COMFORT, Instrumentation Science & Technology, 40:4, 290-304, DOI: 10.1080/10739149.2012.673193

To link to this article: http://dx.doi.org/10.1080/10739149.2012.673193
This article describes the design and implementation of a wearable, multiparameter physiological monitoring system called the Sensing Belt system, which consists of multiple sensors integrated into fabric that communicates with a physiological data acquisition unit (PDAU) that in turn transmits these data to a remote monitoring center (RMC) for analysis. A number of vital signs can be acquired by the system, including electrocardiography (ECG), respiratory inductance plethysmograph (RIP), posture/activity, multipoint skin temperature ($T_{SK}$), and rectal temperature ($T_{RC}$). The physiological data can be stored on a MicroSD card or transmitted to the RMC, where specialized analysis will be provided to extract parameters such as heart rate (HR), respiratory rate (RR), respiratory sinus arrhythmia (RSA), and human energy expenditure. The RMC can receive physiological data from up to 16 Sensing Belt users simultaneously. A medical validation test was carried out to compare the accuracy of the physiological data obtained from the Sensing Belt system with data obtained concurrently from traditional, calibrated laboratory physiological monitoring instruments. The results showed that most of the variables measured by the Sensing Belt are within acceptable error limits. The mean temperature on two trials (walking and running) showed significantly higher mean differences than on other trials, but the correlation coefficient ($r$) remained high (0.985 and 0.989, respectively). This study demonstrates the accuracy of the Sensing Belt system for the monitoring of these physiological parameters and suggests that it could be used to provide a complete human physiological monitoring platform for the study of human heat stress, cold stress, and thermal comfort.

**Keywords** heat stress, physiological monitoring system, sensing belt, thermal comfort, wearable
INTRODUCTION

To study human heat stress, cold stress, and thermal comfort, the researcher must monitor multiple vital signs relevant to physical stress, such as heart rate (HR), heart rate variability (HRV), respiratory rate (RR), skin temperature (T_{SK}), and rectal temperature (T_{RC}). At present, to assess human physiological signals, many separate devices are used to acquire different physiological parameters. Many electrodes, wires, and portable devices must be placed on the human body and hard-wired to one or several data acquisition units to record the physiological data.\textsuperscript{[1]} This method has several disadvantages:

1. The system is complicated to operate. Too many electrodes and wires need to be put on the subjects.
2. The devices are uncomfortable to wear.
3. The large number of devices may produce interference or errors.
4. Because each device needs to upload its data separately, the data processing workload is large.
5. The subject endures some mental and physiological stress from wearing multiple devices, which may affect the results of the experiment.

Recently, new devices with integrated wearable sensors were used in research on heat stress and thermal comfort. Buller et al. used Warfighter Physiological Status Monitoring Program (WPSM) to obtain HR and skin temperature to provide a real-time heat strain risk classifier.\textsuperscript{[2]} Coca et al. used LifeShirt to study the physiological responses of subjects wearing firefighter ensembles.\textsuperscript{[3]} Wearable physiological monitoring systems have become an obvious trend in the study of human physiological responses. A number of wearable physiological monitoring systems have been developed. The first is a monitoring garment used for measuring the electrical activity signals of the heart during daily activities, called the “cross-type” sensing-wise garment, which can stabilize the electrode position and acquire high-quality electrocardiography (ECG) signals.\textsuperscript{[4]} The second, a highly integrated wearable monitoring system targeting sportsmen, measures ECG signals and body impedances with two dry electrodes connected by only one unshielded wire.\textsuperscript{[5]} Finally, a wearable cardiopulmonary monitoring system, the Sensing Shirt, can acquire a number of vital signs, but it uses conventional electrodes to acquire the ECG.\textsuperscript{[6]}

A number of wearable physiological monitoring systems have been put into practical use for health monitoring and ergonomics studies. Vivometrics (bankrupt in 2009) developed a wearable system called LifeShirt, which is capable of acquiring a number of physiological parameters. The system uses adhesive gel-coated carbon electrodes for ECG
sensing and requires these electrodes to be manually affixed to the correct locations on the body. The VitalSense (Respironics) is a telemetric system capable of collecting core body temperature, dermal temperature, HR, and RR data. However, the physiological data can only be viewed on a local VitalSense Monitor, and cannot be viewed in real time at a remote location. Most relevant to our work are the BioHarness (Zephyr) and Equivital (Hidalgo) systems. These two systems are portable physiological data loggers and telemetry systems. However, because only the chest $T_{SK}$ is recorded, they are not suitable for research requiring multipoint $T_{SK}$, such as the study of human thermal comfort.

In contrast with the systems described above, not only can our Sensing Belt system monitor multiparameter physiological signals including ECG, respiratory inductance plethysmograph (RIP), activity, multipoint $T_{SK}$, and $T_{RC}$, but it can also provide real-time viewing and analysis. In our literature survey, we determined that there were no data regarding a system that could acquire multiparameter vital signs, including multipoint $T_{SK}$ and $T_{RC}$, with a highly integrated wearable device. Our system can also provide real-time physiological data viewing and processing for up to 16 users in the RMC.

This article describes the Sensing Belt system, a wearable multiparameter physiological monitoring system that primarily consists of three parts: a wearable Sensing Belt with various integrated sensors; a physiologi-
cal data acquisition unit (PDAU) carried in a pouch attached to the girdle worn around the wrist providing physiological data acquisition, memorizing, analysis, and wireless transmission; and a RMC, where the physiological data will be received, displayed, stored, and analyzed. The overall architecture of the system is illustrated in Figure 1.

METHODS

Sensing Belt

Figure 2A illustrates the Sensing Belt with integrated sensors. The Sensing Belt is designed to be worn underneath clothing, so a comfortable blend of cotton and knitted Lycra was chosen to fabricate the belt. The Sensing Belt consists of an inelastic part, an elastic part, and two Velcro buttons. The elastic part consists of a sinusoidal array of insulated wires embedded in elastic bands woven into the belt to sense respiratory movement. The inelastic part was designed to come in three sizes—small, medium, and large—to ensure there is a size for everyone. The fabric ECG electrodes, the RIP wires, the accelerometer, the chest skin temperature ($T_{SK, CH}$) sensor, and the back skin temperature ($T_{SK, BA}$) sensor are integrated in the fabric of the Sensing Belt and are connected to the PDAU by wires. Three fabric ECG electrodes are stitched into the inner surface of the belt at regular distances along a line connecting...
the subject’s nipples (Figure 2A). The $T_{SK, CH}$ sensor and the $T_{SK, BA}$ sensor are stitched in the front and back positions, respectively, of the inner surface of the belt. The other five temperature sensors—upper arm skin temperature ($T_{SK, UA}$), anterior thigh skin temperature ($T_{SK, AT}$), anterior leg skin temperature ($T_{SK, AL}$), big toe skin temperature ($T_{SK, BT}$), and rectal temperature ($T_{RC}$)—are stitched in the inner surface of the Lycra straps with Velcro buttons, which hold the sensors in position. These sensors are connected to the PDAU by anti-distort wires (Figure 2B). As Figure 2C shows, excluding the $T_{SK, CH}$ and $T_{SK, BA}$ sensors, each of the other four temperature sensors is embedded in a cylindrical protective shell ($\Phi 2 \text{ mm} \times 12 \text{ mm}$) with two wings. The wings have several needle holes ($\Phi 0.5 \text{ mm}$), allowing the protective shells to be stitched onto the straps. A YSI401 (Nikkiso-YSI Co., Ltd, www.nikkiso.com) reusable temperature probe is used as the $T_{RC}$ sensor.

**Physiological Data Acquisition Unit (PDAU)**

Figure 3 shows a block diagram of the PDAU, which includes complete electronic components for signal conditioning, digitization, analysis,
recording, and wireless transmission. The PDAU consists of three parts: an analog section, a digital section, and a power section. The analog section includes signal conditioning circuits for the ECG, RIP, and seven-channel temperature signals. The digital section includes an Mico Control Unit (MCU), C8051F410 (Silicon Laboratories Inc., www.silabs.com), a 4 GB MicroSD card, and a 24-bit high throughput analog-to-digital converter (ADC; AD7732, Analog Devices, www.analog.com). The seven-channel temperature signals are sampled at 50 Hz and are converted to digital signals by the AD7732. The 12-bit ADC embedded in the MCU performs AD conversion for ECG and RIP signals with sample rates of 500 Hz and 50 Hz, respectively. The three-axis acceleration signals from the ADXL345 (Analog Devices) are accessible through a three-wire synchronous serial interface Serial Peripheral Interface (SPI) with the MCU. The analog multiplexers and exciting current direction switches, which act as temperature channel selection and current reversal, respectively, are both controlled by the MCU. The MCU has one universal asynchronous receiver/transmitter (UART) port designed to interface with a wireless Wi-Fi module, HLK-WIFI-M03 (Hi-Link, www.hlktech.com). The system is powered by a 3.7 V rechargeable Li Polymer battery with a capacity of 1.8 Ah. Two low dropout regulators TC1185-3.3 (Microchip, www.microchip.com) are used to generate two separate 3.3 V DC feeds to the analog and digital circuits. A constant-current/constant-voltage linear charger chip, CN3052A (Consonance Electronics, www.consonance-elec.com), powered by USB bus or 5 V DC, is used to charge the Li battery with a programmable charge current of up to 500 mA.

**Sensing Unit**

A customized fabric electrode was developed using electroless metal-plated yarns, which have several useful characteristics such as good conductivity, free tailor, and easy sewing. The major drawback of the fabric electrode is the inherent high skin–electrode impedance and the variance of the contact impedance, which will incur movement artifact noise and add baseline wander. If the input impedance of the instrument amplifier (IA) is not much larger than that of the skin contact, the source signal from the body surface will be very weak and unbalanced. To solve these problems, we used active electrodes built with unit gain buffers embedded in the fabric electrodes. The operational amplifier (OP) OPA129 (TI) was used for its ultrahigh-input impedance. The INA326 set with a gain of five was used as the preamplifier for its low noise and high common–mode rejection ratio (CMRR). To suppress differential DC voltage and offsets that may develop from the half-cell overpotential of the electrodes, an integrator was tied to the IA in feedback to create
a high-pass filter with a corner frequency of 0.5 Hz. The DC voltage at the output of the IA is inverted and gained by the integrator and injected back to null the output. With the DC common-mode voltage removed, the primary gain stage can amplify the ECG signal without becoming saturated. The second stage is a non-inverting amplifier set at $G = 200$ and a second-order Bessel low-pass filter with a $-3 \, \text{dB}$ cutoff at 75 Hz implemented with a low power OP, OPA2335 (TI). The Bessel filter was selected for its linear phase response. In addition, a driven-right-leg circuit is used for powerline interference rejection. The circuit sends inverted common-mode interference present in the circuit to the patient to minimize the noise present on the body.

The system incorporates our patented RIP with the pulse amplitude modulation (PAM) technique to detect respiration with high signal quality and ultralow power consumption.\(^7\) In the system, a constant-current source generates a 15 mA peak-to-peak sinusoidal current at 400 kHz, and the RIP sensor is switched on and off at a frequency of 50 Hz, with an excitation duration of 100 $\mu$s. Although the momentary current through the sensor is very high in order to improve SNR and sensitivity, its average current is only $1/10,000$ of 15 mA. Thus, the PAM can dramatically reduce the power consumption and still keep a high SNR. The RIP signal is sampled by the 12-bit ADC of the C8051F410 MCU at the end of the excitation with a frequency of 50 Hz.

To improve the accuracy of the temperature measurements, a high-precision, multichannel temperature-measuring circuit based on six four-wire B/3-grade PT100 Resistance Temperature Detector (RTDs) and one YSI401 thermistor probe was proposed (Figure 4). The circuit consists of a constant-current source, a current direction switch, voltage reference, analog multiplexers, and an amplifier. The INA326 (TI, www.ti.com) was selected as the amplifier for its high input impedance, low offset voltage ($V_{\text{OS}}$), and low input bias current ($I_B$). To improve measurement accuracy, the current reversal technique was used. The current reversal technique requires the thermoelectric electromotive force ($V_{\text{EMF}}$) to be constant in each measurement phase in order to remove the $V_{\text{EMF}}$ noise contribution. So a high frequency of 50 Hz is used to sample the analog temperature signal; at the same time, the constant-current source reverses at the same frequency. The 20 ms duration between the two measurements is short enough to keep the $V_{\text{EMF}}$ constant, so the $V_{\text{EMF}}$ will be eliminated properly. The multi-axis accelerometer has been proven useful for monitoring human movements, particularly for free-living subjects. An ultralow power, three-axis digital accelerometer, ADXL345 (Analog Devices), integrated in the back of an ECG fabric electrode, is used to capture posture and motility information. The ADXL345 features small size and high resolution (4 mg/LSB). Each axis signal is sampled at a frequency of 200 Hz.
Wireless Communication Architecture

The main task of the wireless communication architecture is to transmit physiological data measured from the Sensing Belt to the RMC. In our system, we used a small size (60 × 24 × 4 mm) Wi-Fi module, HLK-WIFI-M03, to build short-range wireless communication. The module operates in the 2.4 GHz ISM band with a receiver sensitivity of −86 dBm. A maximum range over line-of-sight communication of 200 m outdoors can be achieved using an antenna gain of 2.0 dBi. The physiological data acquired by the PDAU is wirelessly transmitted to the RMC in packets of 79 bytes with a frequency of 25 packets/second. The RMC uses a 150 M WLAN router, TL-WR740 N (TP-LINK, www.tp-link.com), which interfaces with a PC or laptop through an Ethernet cable. Specialized software developed with Borland Delphi 2009 receives and checks the validity of the physiological data from the PDAU; it then performs real-time viewing, archiving, analysis, alarms, and retrospective viewing. The ECG, RIP, and three-axis acceleration signals are displayed continuously in wave form. Seven-channel temperatures are displayed as a data table and real-time trend charts. Other parameters such as HR, RR, and respiratory sinus arrhythmia (RSA) are analyzed and displayed in real time. Data can be exported as comma-separated values (CSV) and binary files for further manipulation in a spreadsheet or database application.

**FIGURE 4** Circuit diagram of seven-channel high-precision temperature measurement. (color figure available online.)
Signal Processing

At the RMC, raw physiological data are subjected to automatic processing, including filtering noise and interference, parameter extraction, finding trends, and generating reports. A subtraction procedure is applied to remove power-line interference; this procedure is very simple and performs well.\[8\] A simple high-pass filter with QRS complex elimination was applied to remove baseline drift. This filter applies a fast linear-phase low-pass filter to extract the baseline drift and then subtracts it from the original data.\[9\] Then, the QRS complex is detected with first-derivative algorithms.\[10\] An extremum searching algorithm is used to identify the critical points of respiratory waves and then to detect the inspiratory and expiratory cycles, as well as to calculate RR. The peak-valley method is used to calculate the RSA. Each axis of acceleration sampled at 200 Hz is passed through a median filter of 13 samples to remove the noise. Then, the energy expenditure can be calculated according to the algorithms described in Bouten et al.\[11\] In the current design, only HR, RR, and RSA are calculated in real time; other parameters such as HRV, QRS width, amplitude, and ST segment can be acquired by off-line analysis.

Validation Test and Practical Trial

A medical validation test was carried out to compare the accuracy of the physiological data obtained from the Sensing Belt worn by test subjects with data obtained concurrently from traditional, calibrated laboratory physiological monitoring instruments. Six male subjects between the ages of 22 and 33 years with moderate to good fitness levels completed four experimental trials with the following protocol: 15 min static standing, 15 min static lying, 15 min treadmill walking at 5 km/h, and 15 min treadmill running at 10 km/h. The subjects’ characteristics were as follows (mean ±SD): age: 26.9 ± 4.2 years; height: 1.71 ± 1.3 m; body mass: 62.6 ± 4.5 kg. The ambient temperature and relative humidity were set at 25°C and 40%, respectively, throughout the test. The study was approved by the institute’s ethics committee, and both oral and written consent were obtained from all subjects prior to their participation in this study. During the test, physiological data were simultaneously acquired from the Sensing Belt and from other instruments. The measured variables and their corresponding instruments included the following: HR (HxM HR Monitor, Zephyr, www.zephyr-technology.com), RR (Piezo Respiratory Transducer MLT1132, ADInstruments, www.adinstruments.com), and skin and rectal temperature (ADInstruments temperature measuring system, including MLT422/D TSK probe, MLT1407 TRC probe, FE228 octal bridge amplifiers, ML312 T-type Pod, and ML880 16/30 channels data acquisition system). The Sensing Belt
and the HxM were worn simultaneously at nearly the same position below the subject’s nipples. The HxM wirelessly transmitted the R-R intervals measured from the ECG to the PC via Bluetooth, where data were recorded as CSV files. Before the test began, the Sensing Belt and ADInstruments temperature measuring system were calibrated to an acceptable accuracy: bias was less than 0.08°C, and $r$ was greater than 0.999 compared with the reference mercury thermometer. Then, multiple temperature probes of the two systems were applied to the subjects. The subjects placed the $T_{RC}$ probes 10 cm into the external anal sphincter, in privacy. The six $T_{SK}$ probes of the Sensing Belt were fixed at corresponding positions with Lycra straps, while six MLT422/D probes were affixed to the same positions using medical-grade tape. An additional strip of tape was used to secure the wiring of the probes 5 cm from the head of the sensors to minimize sensor movement and prevent detachment. Six MLT422/D probes and one MLT1407 probe were connected to the FE228 and ML312, respectively, and then connected to the I2C port and Pod port of MT880. Both respiration and temperature data were recorded and analyzed by specialized software (LabChart 7.2, ADInstruments). All data were transmitted in real time to a PC for storage and later manipulation. The HR, RR, $T_{SK}$, and $T_{RC}$ were analyzed statistically, and absolute agreement between the methods was assessed by determining the mean difference (bias), the standard deviation (SD), the standard error of estimate (SE), the limits of agreement (LoA), and $r$ as described by Bland and Altman.\[12\]

**RESULTS**

The Sensing Belt system was developed and tested for functionality and comfort levels for the users. The PDAU weighs 151 g, with dimensions of 105 mm $\times$ 64 mm $\times$ 24 mm. During the tests, only the last 5 min of ECG and respiration data of each trial were used to calculate HR and RR and were averaged every minute. The Bland–Altman plot for the HR during the four trials is illustrated in Figure 5. Most average HR data resided within the mean $\pm 1.96$ SD range, which signifies good stability in HR detection. Only three data points fell outside the mean $\pm 1.96$ SD range; these outliers were caused by motion interference induced by high speed exercise. Leger and Thivierge have given validity criteria for the tested HR monitor: the monitor was considered to be valid if $r < 0.90$ and SE $> 5$ beats/min.\[13\] Overall, the Sensing Belt HR measurements were valid when compared with the HxM Monitor, with bias $= 0.11$, SE $= 0.16$, and $r = 0.99$ ($n = 72$). Figure 6 shows the Bland–Altman plot for the RR with bias $= 0.25$, SE $= 0.1$, and $r = 0.98$. The results showed that there was a good correlation ($r > 0.94$) for the RR between the two systems.
FIGURE 5 The Bland–Altman plot for the HR from six subjects during four trials. (color figure available online.)

FIGURE 6 The Bland–Altman plot for the RR from six subjects during four trials. (color figure available online.)
The responses of the Sensing Belt and the ADInstruments system showed similar differences across all $T_{SK}$ and $T_{RC}$, regardless of exercise intensity; therefore, only mean temperatures were calculated. In the absence of a “gold standard” for human $T_{SK}$ and $T_{RC}$ measurements, absolute values were presented with no directionality. During standing and lying, the mean differences of the temperature measurements between the two systems were less than 0.5°C. Nevertheless, the Sensing Belt always reported much higher (>0.5°C) mean temperature measurements compared with the ADInstruments system throughout walking and running trials. In the absence of a true reference standard for human skin and rectal temperature measurements, acceptable error limits for the comparison between the two methods were adopted, which were similar to other relevant studies: bias <0.5°C, 1.96 SD < 0.9°C, abs(LoA) < 1.5°C, and $r > 0.9$. Although the mean differences between temperature measurements varied substantially during walking and running trials (0.597, 0.796°C), $r$ remained high (>0.9) and 1.96 SD remained low (<0.9°C), as did the LoA (within ±1.5°C). Furthermore, the differences were uniform and parallel throughout each trial.

**DISCUSSION**

The results of the validation test demonstrated high correlations between physiological measurements of HR, RR, and multipoint $T_{SK}$ and $T_{RC}$ captured concurrently by the Sensing Belt system and calibrated standard physiological monitoring instruments from six subjects while the subjects underwent four different exercise tests. The validity of the HR measurements was similar to the results of other studies published in this area. Compared to the present study, these previous studies also found lower validity coefficients at higher speeds. In summary, the Sensing Belt had high precision in determining HR at the lower-intensity exercises of standing, lying, and walking. During the higher-intensity exercise of running, however, some values were located outside of the confidence interval. When using validity criteria of SE < 5 beats and $r > 0.9$, the Sensing Belt was found to be a valid practical method under common exercise conditions. The high degree of correlation ($r = 0.99$), low bias (0.11), and low SE (0.16) indicate that the Sensing Belt is suitable for a wide range of HR monitoring applications.

In the field of exercise physiology, RR is a vital parameter that can be used to provide information on a person’s physiological state and fitness. In the present study, we acquired chest respiration using the RIP technology, which had significant advantages compared with a standard pneumotachograph for determination of RR, such as better sensitivity, reliability, and security. As expected, a high correlation ($r = 0.98$) was obtained for RR detection by the Sensing Belt and the standard laboratory respiration measuring device.
The results were similar to other relevant studies that compared the accuracy of LifeShirt (which also applies RIP technology) and pneumotachograph recordings of RR in healthy young males at various levels of energy expenditure. These studies ultimately found no significant differences in data from either system.\cite{17,18} The RSA was an important parameter for the study on human temperature regulation mechanism. As there is no standardized test procedure or simulator to evaluate the RSA, more future efforts need to be done on the methodology and clinical validation.

$T_{SK}$ is a fundamental variable in human thermal comfort physiology, and yet $T_{SK}$ measurement remains impractical in most free-living experiments using currently available methods. We designed a high-precision, seven-channel temperature measuring circuit using current-reversal technology. With calibration, a high accuracy was achieved (bias $<0.08^\circ$C, $r>0.999$). A comparison of measurements of the mean temperature during four trials on humans taken by the Sensing Belt and the ADInstruments system demonstrated that both systems could provide temperature measurements within acceptable error limits. However, the Sensing Belt consistently reported a significantly higher mean temperature than the ADInstruments system throughout walking and running trials, despite the fact that both systems had been calibrated. The observed differences between the two systems could be attributed to some differences in microenvironments caused by contact thermometry.\cite{19,20} A microenvironment between strap and skin may exist that affects the sensor to read consistently higher temperatures throughout the trials, although a good air-permeability Lycra strap has been selected to fix temperature sensor in our design. Another possibility for the lower mean temperature using the ADInstruments system was that the ADInstruments' probe had a smaller contact surface area, and its wired attachment increased the likelihood of the probe losing direct contact with the skin, which might cause the temperature measurements to tend toward the ambient temperature and give false results. Although precautionary measures were taken to prevent the probes from changing position (medical adhesive tape was used to fix the probes and wires), the probes still lost contact with the skin, which is a major disadvantage of all hard-wired temperature measurement devices. In our design, the probe was sewn onto a Lycra strap to keep the probe firmly against the skin, so the temperature measurements of the Sensing Belt were independent of the temperature of the ambient environment. Fortunately, these mean temperature differences were parallel and consistent, and the $r$ remained high under all trials, so these temperature differences can be easily accounted and adjusted for.

Acceleration data were acquired, but only motility information and the energy expenditure of the wearer were obtained. However, the acceleration data could be very useful; for example, they could be used as a reference for removing motion artifacts. Furthermore, accelerometer sensors have been used
to assess the core body temperature of soldiers exposed to extreme cold by analyzing shivering characteristics.\[^{21}\] Our future work will further analyze acceleration data to obtain more valuable information, such as detecting hypothermia.

We have made progress in addressing many of the limitations of wearable physiological monitoring systems. The Sensing Belt is fabric-based and allows simultaneous monitoring of multiparameter physiological signals, real-time transmission or logged monitoring mode, and data viewing for up to 16 users. However, there are considerable possibilities for improvement. In the current prototype, only RR and RSA were obtained from the RIP respiration signal. However, tidal volume ($V_T$) and minute ventilation ($V_E$) are very important parameters for applications in occupations where heat stress is serious, such as firefighting, mine rescue, construction, and roofing.\[^{3}\] The Sensing Belt system will have broader applications if it can acquire $V_T$ and $V_E$. Another limitation of our study is the use of an inserted rectal probe to detect core body temperature, which causes some discomfort to the wearer. The need to use an inserted rectal probe is a substantial, but not insurmountable, problem for core temperature acquisition. Our future work will apply more advanced technology, such as an ingestible temperature sensor or a Kalman filter (KF) approach, to noninvasively obtain core temperature.\[^{22,23}\]

**CONCLUSION**

A wearable, multiparameter physiological monitoring system called the Sensing Belt system has been developed by integrating multiple sensors into fabric and acquiring and transmitting physiological data to the RMC using the PDAU. The Sensing Belt system aims to provide a complete human physiological monitoring platform, suitable for both laboratory and field research. In the future, the system will be military-qualified, increase wireless transmission ranges, provide more useful and reliable physiological data, and become more comfortable to wear.

**ACKNOWLEDGMENTS**

This work was supported in part by the National Natural Science Foundation of China (Grant Number: 60801009) and the Beijing Natural Science Foundation (Grant Number: 3102028).

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